BME Horizon Vol 2 Issue 1 2024
DOI: 10.37155/2972–449X–vol2(1)–92

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Investigation of Titanium-Based Biomaterials Used in Implant Applications

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Received: December 1, 2023; **Accepted:** March 27, 2024; **Published Online:** March 30, 2024

How to cite: Yılmaz, E., Findik, F. Investigation of titanium-based biomaterials used in implant applications. *BME Horizon*, 2(1). Doi: https://doi.org/10.37155/2972-449X-vol2(1)-92.

Abstract: Biomaterials are natural or synthetic materials used to fulfill or support the functions of living tissues in the human body. Implants, on the other hand, refer to inanimate materials placed inside the body and living tissues. Implants are also used to replace the functions of teeth and other bones in the body. In this study, first, the structure and mechanical properties of 208 bones in the body were examined. Then, the requirements of biomaterials used in implant applications were criticized. Finally, the properties of titanium and its alloys, one of the metallic biomaterials, were investigated, and the focus was on how these alloys were produced using powder metallurgy.

Keywords: Ti-based alloy; Implant; Biomaterial; Porosity; Powder metallurgy; Production; Bone; Mechanical properties

1. Introduction

Biomaterials are natural or artificial materials used in the construction of implants to restore the function and shape of lost or diseased biological structures. They are used in various parts of the human body, including artificial heart valves in the heart, stents in blood vessels, and implants placed in place of teeth, ears, elbows, hips and knees. Among these, the number of implants used for spine, knee and hip is quite high. Causes of joint renewal changes; Diseases such as osteoporosis (bone resorption/ excessive bone porosity), osteoarthritis (inflammation in bone joints) and trauma. According to data collected on total joint replacement surgery, there is a high demand for permanent implants (by the end of 2030). Total hip implants will increase by 174% and total knee implants will increase by 673% [1].

The development of orthopedic implants is a complex and multi-field scientific issue. For the implant material to be successful, it must have mechanical properties

compatible with bone, contain biocompatible elements, have good osseointegration properties and have high corrosion resistance. Co-Cr alloys and stainless steels were first used for load-bearing metallic implant materials due to their low cost and high mechanical properties. Later, these traditional biomaterials were replaced by Titanium (Ti) and its alloys, which have a lower elastic modulus, are more biocompatible and have higher corrosion resistance [2].

Beta (β) type Ti alloys containing niobium (Nb), zirconium (Zr), Tantalum (Ta) have been widely used in implant applications in recent years due to their low elastic modulus and high corrosion resistance [3]. Although β-stable constitutive elements reduce the elastic modulus, the values of bone are not approached. Porous alloy production is preferred to reduce the mechanical incompatibility between Ti alloys and bone. Moreover, the porous structure promotes bone tissue growth into the implant porosity, providing better biological fixation. Depending on age, the porosity rates of cortical bone and cancellous bone are between 5-30% and 70-95%, respectively. Porosities allow nutrient and body fluid transport. Therefore, porosity is important for biological activities and mechanical properties [4].

However, one of the main defects of Ti alloys is the bioinert oxide passivation layer, which can quickly form on the surface of Ti implants under physiological conditions, leading to a poor osteoinductivity of these implants [5]. The success of the Ti implant in vivo relies on a process known as osseointegration, which promotes bone healing and the formation of new bone. The bioinert surface, which prevents the osseointegration process from occurring, often leads to the formation of a fibrous tissue around the implants. For this reason, bioactive coatings such as hydroxyapatite (HA) are commonly applied to the Ti implant surface. HA ($Ca_{10}(PO)_4(OH)_2$) is a widely used biomaterial due to its stability in the physiological environment and chemical similarity with the mineral component of bone [6], [7]. However, its use is limited to low load carrying applications due to its poor mechanical properties. To solve this problem, materials such as polyethylene, Al_2O_3, titanium oxide (TiO_2), Y_2O_3, yttria stabilized zirconia, Ni_3Al and carbon nanotubes have been used in the literature to strengthen the mechanical properties of HA [8].

In this study, first the structure and mechanical properties of 208 bones in the body were examined. Then, the requirements of biomaterials used in implant applications were critisized. Finally, the properties of titanium and its alloys, one of the metallic biomaterials, were investigated, and the focus was on how these alloys were produced using powder metallurgy.

2. Characteristics of Bone

2.1 Content and Structure of Bone

Bone is a living and dynamic structure. There are 208 bones in the human body. It is one of the hardest tissues in the body. They perform many important functions; they protect internal organs, help body movement, can release calcium and phosphate ions into the bloodstream, the formation of blood cells occurs in the marrow cavities of long bones. The content of bone is basically divided into cells and intercellular substances (**Figure 1**.).

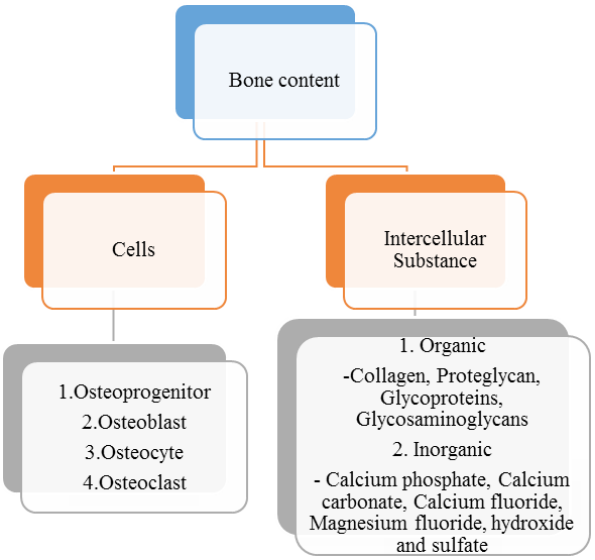

Figure 1. Contents of bone.

The appearance of bone cells is given in **Figure 2**. Osteoprogenitor cells are also called osteogenic cells, they are mesenchyme cells conditioned to become bone cells and have the properties of stem cells. In mature bone, they are found inactive in the membranes surrounding the bone (periosteum, endost) and around the vessels in the canals (haversian and volkmen canals). They resemble fibroblasts in shape. During new bone formation and repair of bone fractures, they multiply by mitosis and turn into osteoblasts.

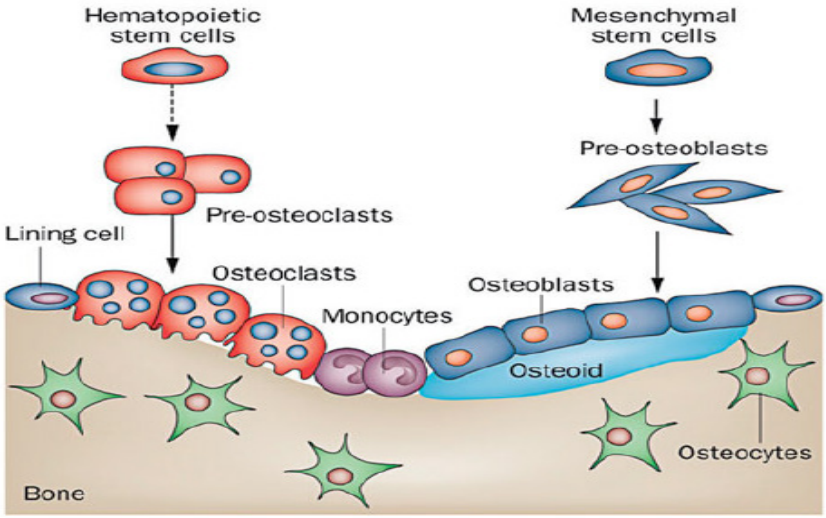

Figure. 2. Bone cells [9].

Osteoblasts are bone-forming cells, arranged in a single row on the surfaces of spongy bone (trabeculae) or compact bone (lamellar). They secrete the organic part of the bone matrix (collagen, proteoglycan, etc.), such basic substance that has not yet calcified is called osteoid. Osteoblasts remain embedded in the osteoid tissue, and when this tissue calcifies, their activity decreases, their shape becomes flattened, and they become osteocytes. Calcification is achieved by the precipitation of calcium and phosphorus ions from the blood by the secretion of alkaline phosphate enzyme by osteoblats. Osteoblasts trapped in the secretion they produce turn into osteocytes. Osteocytes are mature bone cells. They have a flat oval shape and are in cavities called lacun. Since the bone matrix has hardened, material transport occurs through osteocytes. This is why these cells have long extensions and these extensions are connected to each other. They ensure the nutrition of the bone tissue. If they grow old and die, the bone tissue in that area cannot be nourished and is absorbed by osteoclasts, and new bone tissue is made instead. Osteoclasts are cells that provide bone resorption (destruction). Therefore, they appear during bone destruction and then disappear. They are giant cells formed by the union of monocytes coming from the blood. The places where osteoclasts sit on the bone surface become hollow due to melting with long and short cytoplasmic extensions like saw teeth (**Figure 2.2**.).

Like other support tissues, intercellular substance is abundant in bone tissue, and cells are in the minority. The inorganic part constitutes 30-40% of bone tissue. 85% of this is calcium phosphate, 10% is calcium carbonate, 5% is calcium fluoride, magnesium fluoride, hydroxide and sulfate compounds, sodium, and potassium. Calcium and phosphate ions in bone are in the form of hydroxyapatite crystals. Water and other ions collect in a thin layer on their surface. Hydroxyapatite is the substance that gives hardness to bones. The organic part of bone tissue is synthesized and secreted by osteoblasts, consists of amorphous and fibrous elements (collagen, etc.) like other supporting tissues, and constitutes 60-70% of bone tissue. Collagen fibers are arranged parallel to each other and HA crystals are located between them. Collagen is the structure that ensures that bones are hard but do not break easily.

Bones morphologically consist of two parts: the outer part is compact (cortical) and the inner part is spongy bone (**Figure 3**.). When viewed with the naked eye, cortical bone tissue appears homogeneous and compact. However, when examined under a microscope, it is seen that it is equipped with small canals (Havers and Wolkman canals). In compact bone, cells take up little space and dominate the matrix structure. The channels running parallel to the long axis of the bone are called Haversian channels (osteon), and the channels connecting them are called Wolkman channels. Blood vessels and nerves run through both. Bone tissue is fed by the diffusion of nutrients coming out of these channels through osteocytes. There are lamellae next to the channels and spaces called lacuns

between the lamellae. There are osteoblasts in these spaces, which then turn into osteocytes. There are no osteons in spongy bone; it consists of lamellae and osteocytes arranged irregularly by canals. Nutrients and oxygen are taken from the bone marrow. The bone membrane covers the outer bone surface, and the inner bone membrane covers the inner bone surface. They consist of the cells and connective tissue that form the bone. There is an epiphyseal growth plate at the end of the long bone, except in adults. Epiphyseal growth plate; separates the bone tip and body. When growth is completed, this area ossifies and disappears. The surface attachment of the bone end is covered by articular cartilage [10], [11].

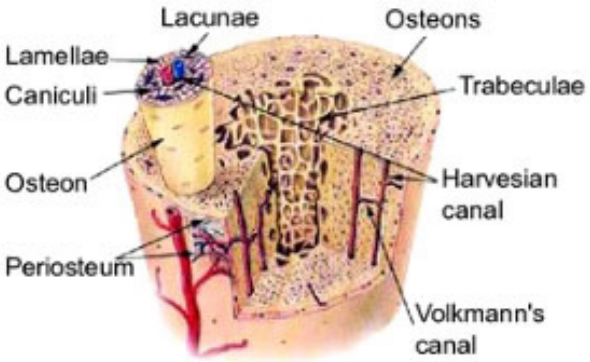

Figure 3. Natural bone structure [12].

2.2 Mechanical Properties of Bone

The strength of the bone varies depending on factors such as the direction of the applied load, gender, age, and calcium content. The most important difference between cortical and cancellous bone is the porosity ratio. While the porosity of cortical bone increases from 1% to 30% with increasing human age, this rate increases from 70% to 95% in cancellous bone. While the density of cortical bone is 1.85 g/cm^3, the density of cancellous bone is 0.30-0.60 g/cm^3. As porosity increases, strength and density values decrease. The mechanical properties of cortical and cancellous bone are summarized in **Table 1**. The strength of bone in compression is higher than in tension. Small increases in percent mineralization result in large increases in both elastic modulus and strength. For example, for cortical bone, when the calcium (Ca) content was approximately 300 mg/g, the elastic modulus was close to 30 GPa, while when the Ca content decreased to 200 mg/g, the elastic modulus was measured to be approximately 5 GPa. The percent strain of cortical bone under load is between 1-3%, while the percent strain of cancellous bone is between 5-7% [13].

Table 1. Variation of mechanical properties according to bone type [13].

Bone Type	Elastic Modulus (GPa)	Compression Strength (MPa)	Tensile Strength (MPa)	Fracture Toughness (MPa.m$^{1/2}$)
Cortical	7-30	100-230	50-150	2-12
Spongy	0,05-0,5	2-12	10-20	-

3. Biomaterials Used in Implant Applications

3.1 Biomaterial Requirements

The development of orthopedic implants is a complex and multi-field scientific issue. Parameters such as mechanical properties, surface characteristics, material composition and crystal structure determine implant life. The various reasons why implants fail are summarized in **Figure 4**.

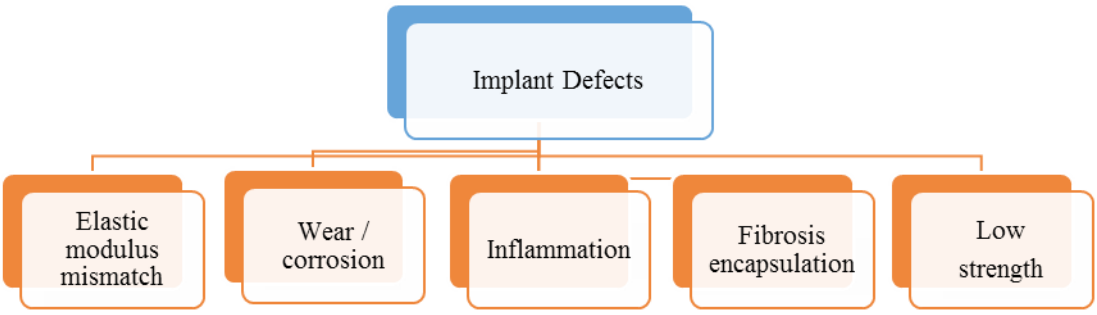

Figure 4. Implant defects [1].

The materials used as implant materials are expected not to cause allergic or inflammatory reactions in the human body and to be non-toxic. The success of biomaterials depends heavily on the human body's response to the implant. **Table 2** gives the classification of biomaterials based on the response of the human body. Bioactive materials are preferred because they combine well with the surrounding bone,

but biotolerant implants can be accepted for implant manufacturing after appropriate surface treatments. When implants are exposed to human tissues and fluids, various reactions occur between the implant material and surrounding tissues. These reactions determine the acceptability of this material by the body system. Issues regarding biocompatibility are as follows; 1) Thrombosis (blood clotting) is blood clotting and adhesion of blood clots to the biomaterial surface. 2) Encapsulation of biomaterials with fibrous tissue. Implant loosening occurs when an implant surface cannot fuse with other tissues and adjacent bone due to micro movements. If the implant is not well bonded with the bone, a fibrous layer forms between the bone and the implant. Therefore, it is important that the implant integrates well with the adjacent bone (osseointegration). Surface chemistry, surface roughness and topography play important roles for good osseointegration. If the implant surface is prepared with bioactive or appropriate surface characteristics, success is achieved in osseoinduction (stimulation of bone formation), osseoconduction (bone development on the surface) and osseointegration (fixation of the implant through bone-implant contact).

Table 2. Classification of biological materials based on their interaction with the surrounding tissue [1]

Classification	Reaction	Examples	Effect
Biotolerant Materials	Formation of thin connective tissue capsules (0.1-10µm), the capsule does not adhere to the implant surface.	Polymer-poly tetra fluoromethylene (PTFE), polymethyl methacrylate (PMMA), Ti, Co-Cr, etc.	Rejection of the implant leads to implant failure
Bioactive Materials	Bone tissue formation around the implant integrates strongly with the implant surface.	Bioglass, synthetic calcium phosphate containing hydroxyapatite (HAP)	Acceptance of the implant ensures successful implantation.
Bioreabsorble Materials	Replacement with autologous tissue	Polylactic acid, polyglycolic polymer, engineered bone grafts, whole tissue bases or protein and structural support system composites,	Acceptance of the implant ensures successful implantation.

If implant fractures are caused by insufficient strength or mechanical property mismatch between bone and implant, then this material is biomechanically incompatible. The elastic modulus of the material replaced with bone is expected to be equivalent to bone. Depending on the measurement direction and type of bone, the elastic modulus varies between 4-30GPa [4]. Current implant materials are harder (rigid) than bone, preventing the necessary stress from being transferred to the adjacent bone, bone resorption occurs around the implant, causing implant loosening. The biomechanical mismatch that leads to the death of these bone cells is defined as the 'stress shielding effect' [14]. Therefore, an implant with high strength and close to the elastic modulus of the bone is required to ensure a high service life and to prevent implant loosening. Additionally, because of the low wear and corrosion resistance of implants in body fluid, metal ions that are incompatible with the body are released from the implant. The released ions have been found to cause allergic and toxic reactions. The service life of the material is determined mainly by its wear resistance. In addition, low wear resistance causes implant loosening, and wear debris accumulates in the tissue and causes various reactions. Therefore, high corrosion and wear resistance is of primary importance for the long life of the material in the human system.

3.2 Metallic Biomaterials

Metals are widely used in joint replacements such as internal support, biological tissue and stents, orthopedic fixation, and tooth roots. The metals and alloys primarily used in biomedical applications are stainless steels, Co alloys and Ti alloys. These materials find use in replacing hard tissues such as hips and knees, especially due to their high reliability in terms of mechanical performance. Some mechanical properties of metallic biomaterials in comparison with bone are given in **Table 3**. The elastic modulus of Co-Cr alloys and stainless steel is approximately 10 times that of bone, leading to the 'stress shielding' effect. The elastic modulus of Ti and its alloys is approximately half of that of stainless steel, therefore the risk of 'stress shielding' of Ti and its alloys is less compared to other biomaterials.

Table 3. Comparison of mechanical properties of metallic biomaterials with bone [15].

Materials	Elastic Modulus (GPa)	Yield Strength (MPa)	Tensile Strength (MPa)	Fatigue Limit(MPa)
Stainless steel	190	221-1213	586-1351	241-820
Co-Cr allloys	210-253	448-1606	655-1896	207-950
Ti	110-120	485	760	300
Ti-6Al-4V	116-120	896-1034	965-1103	620
Cortical bone	15-30	30-70	70-150	-

SUS 316L austenitic stainless steel is the only reported stainless steel used in the biomedical field. However, it has been found that these Ni-containing alloys show allergic reactions. Additionally, pitting, cracking, and stress corrosion have been reported for implants manufactured from SUS 316L. To prevent allergic reactions to Ni, an austenitic stainless steel with a high nitrogen content has been developed. Therefore, the new research trend is to develop Ni-free stainless steel.

The wear resistance of Co alloys is higher than both Ti alloys and stainless-steel alloys. In artificial hip joints, the joint head is subject to wear. Therefore, hip joints are manufactured from Co-Cr-Mo alloys that exhibit high strength and ductility. It has been reported that the distribution of carbides in Co alloys increases the wear resistance of these alloys. Additionally, the formation of a martensitic phase by deformation stimulation improved the wear resistance of Co alloys [16].

The high biocompatibility of Ti and its alloys has led to their preferential use in medical and dental fields compared to other alloy systems. The properties that make Ti alloys the most important metallic material in the biomedical field are as follows: They have good mechanical properties, excellent corrosion resistance due to the TiO_2 solid oxide layer, good biocompatibility, relatively low elastic modulus, low weight and non-magnetic behavior. The above-mentioned properties make Ti and Ti alloys preferred options for implantation. However, its weak tribological properties and bioinert surface are its disadvantages. A comparison of metallic biomaterials used in the human body is made in **Table 4**. Accordingly, Ti and its alloys are preferred due to their high biocompatibility, low elastic modulus, high corrosion resistance and low density compared to stainless steel and Co-Cr alloys. In addition, with the development of beta phase-based Ti alloys (containing Nb, Ta, etc.) in recent years, elastic modulus values and other mechanical properties have decreased to lower values than those of pure Ti and Ti6Al4V. Especially with the addition of Nb to Ti alloys, wear and corrosion resistance has improved with the formation of Nb_2O_5 on the surface. In addition, studies are being carried out to improve the bioproperties of metallic implants with bioactive coatings on the Ti alloy surface [17, 18].

Table 4. Comparison of metallic materials used in the human body [15].

Metal and alloys	Selected examples	Advantages	Disadvantages	Applications
Ti and alloys	Pure Ti, TiAlV, TiAlNb, Ti13Nb13Zr, TiMoZrFe	High biocompatibility, low elastic modulus, high corrosion resistance, low density	Low tribological properties, toxic Al-V	Bone and joint replacements, fracture fixation, dental implant, pacemaker encapsulation
Co and Cr alloys	CoCrMo, CrNiCrMo	High wear resistance	Ni allergy, high elastic modulus	Bone and joint replacements, dental implants, dental restorations, heart valves
Stainless steels	316L	High wear resistance	Ni allergy, high elastic modulus	Fracture fixation, stent, surgical instruments
	NiTi	Low elastic modulus	Ni allergy	Bone plates, stents, orthodontic wires
Other	Platinum, Pt-Ir	High corrosion resistance under extreme voltage and load transfer		Electrode
	HgAgSn	Easy shaping in the oral environment	Hg toxicity	Dental restoration

3.2.1 Properties of titanium and its alloys

Although aluminum and magnesium are generally known as light metals, titanium also has a density of approximately 60% of the density of iron (4.51g/

cm^3). Titanium is a special material of interest in both engineering and dental and medical applications. Ti is a transition element, it is in group IV A of the periodic table, its atomic number is 22 and its atomic weight is 47.88. Titanium has low thermal conductivity (21.4 w/m.K at 20-25°C) and electrical conductivity is also low. Also, it is not magnetic. It is divided into 4 classes according to oxygen content (maximum oxygen content 4%). Titanium's affinity for oxygen creates a passive layer on its surface and provides resistance to corrosion. It is widely used in biological applications and does not cause allergic reactions.

Figure 5 shows the Ti-Nb binary phase diagram and relationship of temperature with the mole fraction of Nb. There are four different phase zones in the binary diagram including α, β, α+β and liquid phases. α phase area is very narrow and having very little Nb in the α phase. Whenever Nb included into Ti element, β phase regions are getting larger, while α+β phase zone is getting smaller. Titanium has an allotropic transformation (**Figure 5**). While Ti is in the alpha (α) phase (hexagonal close-packed: HSP) up to 1155 K (882°C), it turns into the β phase (body-centered cubic; BCC) above this temperature. 1943 K (1670°C) is the melting temperature. Additionally, it is seen that beta phase stability increases with the addition of Nb to Ti. Apart from these equilibrium phases, non-equilibrium phases (αı, αıı, ω) may occur depending on alloy ratios and cooling rate variables. αı (hexagonal structured martensite), αıı (orthorhombic structured martensite) phases are formed by rapid cooling from the β phase. The hexagonal omega (ω) phase is formed from the β phase by slow quenching or aging at an average temperature [19]. Among these phases, the phase with the lowest elastic modulus is the β phase, and the phase with the highest elastic modulus is the ω phase [20]. While aluminium, nitrogen, oxygen, and carbon stabilize the α phase, vanadium, molybdenum, niobium, and tantalum stabilize the β phase. The α1 martensite phase can occur with low alloying. When β-stabilizers are present in more than critical amounts, the martensite structure turns into an orthorhombic structure. The critical solute content for the αı/αıı limit depends on the solute elements, in the Ti-Nb alloy the αı phase is 6% at. It occurs up to Nb content. When the amount of β stable constitutive alloying element exceeds a certain value, β Ti is formed (because of quenching or cooling in air). This point is reached with the addition of approximately over 40% Nb (at.) for Ti-Nb phase diagrams (**Figure 5**). In stable alloys, there is no phase transformation by thermomechanical processes, whereas in metastable β; α precipitation or other transformations may occur.

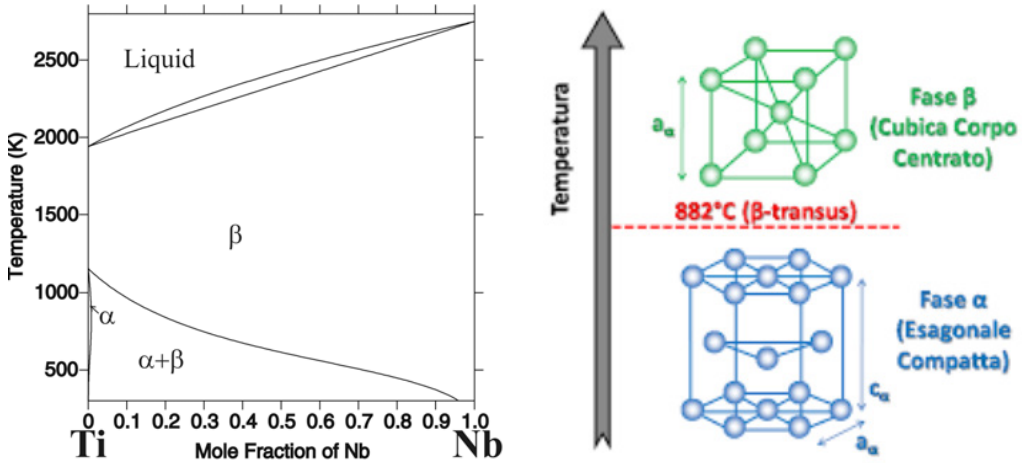

Figure 5. Ti-Nb phase diagram and allotropic transformation of Ti [21].

While pure Ti is commonly used in dental implants, it is inadequate in situations where high strength is required, such as hard tissue replacement (**Table 5**). To overcome this limitation, pure Ti was replaced by α+β type Ti-6Al-4V. This alloy was developed in America in 1954, it is the first Ti alloy used in practice, it has good heat resistance, strength, flexibility, toughness, machinability, weldability, and corrosion resistance. However, toxic effects of Al and V elements have been observed in long-term use. For this reason, since the

1990s, second-generation biocompatible and beta-stable alloys containing constructive elements (Nb, Ta, etc.) have been developed [22]. In this way, Ti alloys with elastic modulus below 100 GPa and superior corrosion resistance and bio properties were obtained (**Table 5**).

Table5. Mechanical properties of biomedical Ti alloys [1].

Ti alloy/microstructure	Elastic modulus (GPa)	Yield strength (MPa)	Tensile strength (MPa)
1st generation			
Commercially pure Ti (α; grade 1-4)	100-105	240-692	785
Ti6Al4V ELI (α+β; standard grade)	112	850-900	895-930
Ti6Al7Nb (α+β; forged)	110	880-950	900-1050
Ti5Al2,5Fe (α+β)	110	895	1020
2nd generation			
Ti13Nb13Zr (β; forged)	79-84	836-908	973-1037
Ti12Mo6Zr2Fe (TMZF; β)	74-85	1000-1060	1060-1100
Ti35Nb7Zr5Ta (TNTZ; β)	55	530	596
Ti29Nb13Ta4,6Zr (β; aged)	65	864	911

3.2.2 Production of titanium and its alloys

Ti-based implant production is a relatively difficult process due to reasons such as the complexity of the products, high cost, and high melting temperatures of the raw materials (Ti, Nb, Zr, Ta, etc.). Titanium and its alloys are suitable for thermoforming and welding. It can be produced by casting or powder metallurgy methods [23]–[24]. The fact that Ti and its alloys have a high melting point, that it reacts and oxidizes with oxygen, nitrogen, and hydrogen in the air when it encounters air, and that it reacts with the ceramic crucible makes it difficult to produce this material with the classical casting method. Ti alloys can be produced in a protective atmosphere by induction melting and electric arc melting methods. However, when alloying elements with different melting temperatures are used (for example, Ta: 3017°C), segregation problems may occur. Therefore, traditional powder metallurgy and powder injection molding (PIM) methods were used in this study. With PIM, a product close to a net shape can be obtained, it does not require any additional post-production plastic shaping, and high-strength and dense products can be obtained by using the appropriate production method and binders. In addition, products with homogeneous and controllable porosity can be obtained by adding spacers or binders in appropriate sizes and proportions using powder metallurgy methods [26]. For implant applications to be used instead of bone, the structure of the bone is imitated with porosity production. In this way, the density of the metallic implant is brought closer to that of cortical bone (1.6-2 g/cm^3). Most importantly, the elastic modulus of the implant is matched with that of the cortical bone (5-30 GPa), preventing the problem of implant loosening. In terms of bioproperties, new bone tissues developing into the pores increase the osseointegration feature. In recent years, various Ti-Nb based alloys have been produced by powder metallurgy, their mechanical and microstructural properties have been investigated and the results have been reported in various journals [27-29]. The chemical compositions of these various investigated alloys are given in **Table 6** and **Table 7**. In addition, the microstructure images of the powders used in the studies before and after sintering are given in Figure 6 and **Figure 7**.

Table 6. Chemical composition of alloys produced with PIM. [30]

Alloy code	Composition (%wt.)			
	Ti	Nb	Sn	Zr
Ti	100	-	-	-
Ti16Nb	84	16	-	-
Ti28Nb	72	28		
Ti40Nb	60	40	-	-
Ti16Nb2Sn	82	16	2	-
Ti16Nb4Sn	80	16	4	-
Ti16Nb5Zr	79	16	-	5
Ti16Nb10Zr	74	16	-	10
Ti16Nb15Zr	69	16	-	15

Table 7. Porous alloy compositions. [30]

Alloy code	Ti(%wt)	Nb(%wt.)	Zr(%wt.)	Ta(%wt.)	Spacer(% volume)
Ti16Nb	84	16	-	-	0
Ti16Nb30SH	84	16	-	-	30
Ti16Nb50SH	84	16	-	-	50
Ti16Nb70SH	84	16	-	-	70
Ti16Nb10Zr	74	16	10	-	50
Ti16Nb10Zr5Ta	69	16	10	5	50
Ti16Nb10Zr10Ta	64	16	10	10	50

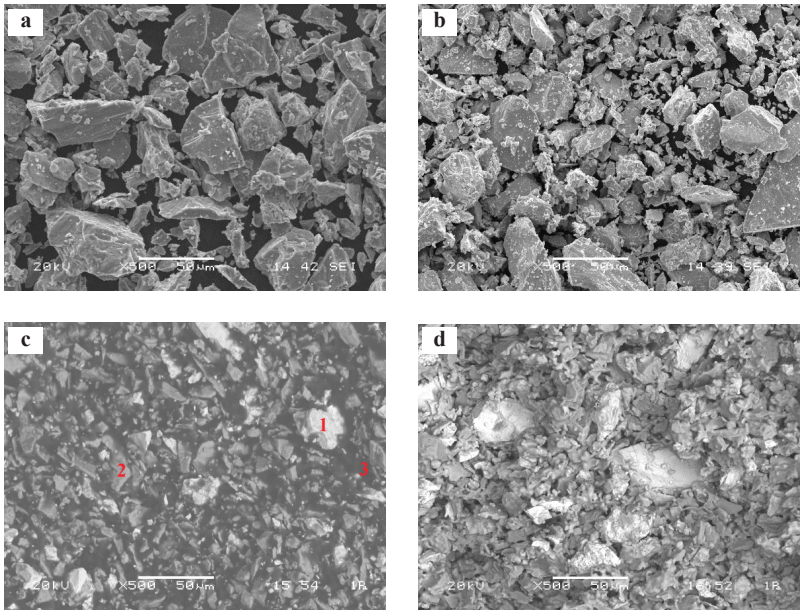

Figure 6. SEM images of **(a)** Ti powder, **(b)** Nb powder, **(c)** Ti40Nb alloy after injection molding, **(d)** Ti40Nb alloy after pre-sintering. [30,31]

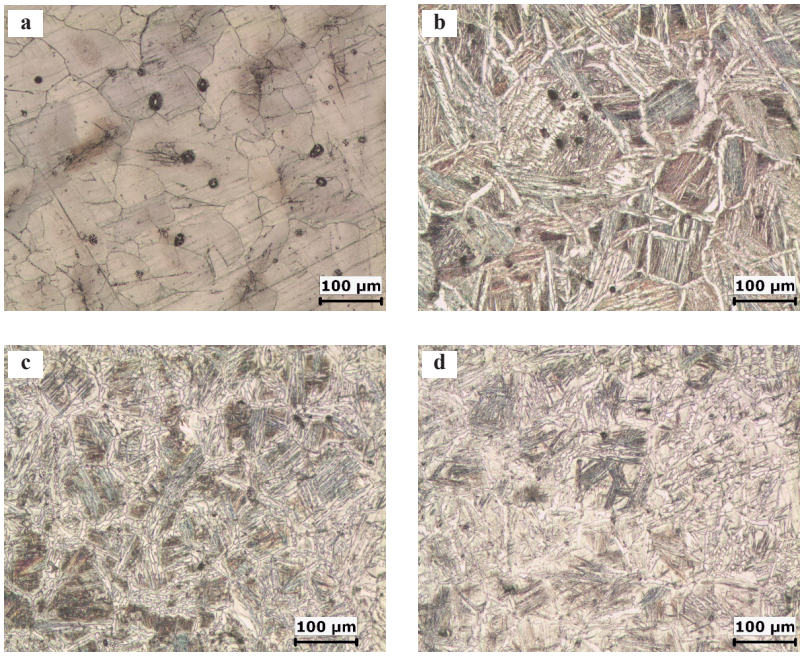

Figure 7. Microstructures of sintered **(a)** Ti, **(b)** Ti16Nb, **(c)** Ti28Nb, **(d)** Ti40Nb alloys. [30,31]

4. Discussion

Metallic materials are commonly preferred for biomaterials used instead of bone. It is desired that the implant material has properties compatible with bone. Titanium alloys have superior properties than Co-Cr alloys and stainless steels in terms of strength / density ratio and elastic modulus value. Titanium is non-toxic and has good corrosion resistance due to the oxide layer on its surface. The compressive strength of Ti6Al4V alloy (1300-1750 MPa), which is widely used in biomaterial applications, is above that of pure titanium (400-1000 MPa). However, with long-term use, a cytotoxic effect occurs due to the presence of Al and V. For this reason, in recent years, interest in Titanium alloys containing β-phase stable forming elements (such as Nb, Ta) instead of Al and V elements has increased. As seen in **Table 8**, compared to Ti-6Al-4V alloy, the compressive strength in Ti-16Nb alloy is approximately the same, but the elastic modulus has decreased by approximately 12% [32-35]. The elastic modulus is still well above that of cortical bone (5-30 GPa). Elastic module incompatibility between the implant material and the surrounding bone tissue results in implant loosening. Therefore, porous production is required to make the elastic modulus value compatible with bone. The elastic modulus of the porous Ti-16Nb alloy with 50% spacer content is compatible with bone. With the production of porous alloy, the density value can be adjusted to the same value as that of bone. In addition, osseointegration property develops with apatite growth into the pores and new bone tissue formation. By adding Nb to Ti, the Nb-containing oxide formed on the surface in SBF is more stable than titanium oxide. In this way, corrosion resistance increases. However, as the surface area increases in the porous structure, corrosion resistance decreases.

Table 8. Comparison of implant materials

Material	Elastic modulus (GPa)	Compressive Strength (MPa)	Density (g/cm^3)	Icorr($\mu A/cm^2$)
Cortical bone	5-30	130-200	1.6-2	-
Pure Ti	110-120	400-1000	4.5	0,79
Ti-6Al-4V	110-114	1300-1750	4.42	1.575
Ti-16Nb	97	1450	4.66	0.275
Porous Ti-16Nb (50 wt.%)	15	100	1.91	2.21

Mechanical properties and density values of porous Ti-Nb alloys are compatible with bone. In recent years, bioactive and antibacterial coatings have been studied to increase corrosion resistance. Porous Ti-Nb alloys with bioactive coating on the surface can be considered as a good candidate material for both orthopedic and dental implants.

5. Conclusions

Biomaterials are natural or synthetic materials used to fulfill or support the functions of living tissues in the human body. The findings obtained in this study are summarized as follows:

- Bone is a living and dynamic structure. There are 208 bones in the human body. It is one of the hardest tissues in the body. They perform many important functions; They protect internal organs and help body movement.
- The strength of the bone varies depending on factors such as the direction of the applied load, gender, age, and calcium content. The most important difference between cortical and cancellous bone is the porosity ratio. While the porosity of cortical bone increases from 1% to 30% with increasing human age, this rate increases from 70% to 95% in cancellous bone. As porosity increases, strength and density values decrease.
- The materials used as implant materials are expected not to cause allergic or inflammatory reactions in the human body and to be non-toxic.
- The metals and alloys primarily used in biomedical applications are stainless steels, Co alloys and Ti alloys. These materials find use in replacing hard tissues such as hips and knees, especially due to their high reliability in terms of mechanical performance.
- Titanium is a special material that is of interest both in engineering applications and in dental and medical applications. Titanium has low thermal conductivity (21.4 w/m.K at 20-25°C) and electrical conductivity is also low. Also it is not magnetic. Titanium's affinity for oxygen creates a passive layer on its surface and provides resistance to corrosion. It is widely used in biological applications and does not cause allergic reactions.
- Ti-based implant production is a relatively difficult process due to reasons such as the complexity of the products, their high cost, and the high melting temperatures of the raw materials (Ti, Nb, Zr, Ta, etc.).

Titanium and its alloys are suitable for thermoforming and welding. It can be produced by casting or powder metallurgy methods. In this study, traditional powder metallurgy and powder injection molding (PIM) methods were used.

- It has been understood that Ti-Nb based alloys are an alternative to Ti-Al-V alloys by examining their mechanical and microstructural properties.

Conflict of Interest

Declaration of conflict of interest.

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