DEVELOPMENT AND CHARACTERIZATION OF BIOMEDICAL POROUS Ti-15Mo ALLOY

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METALLURGICAL AND MATERIALS ENGINEERING

Hussein Basim Baqer OBAIDA

Thesis Advisor
Assoc.Prof.Dr. Hüseyin DEMİRTAŞ

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Hussein Basim Baqer OBAIDA

Thesis Advisor
Assoc.Prof.Dr. Hüseyin DEMIRTAŞ

T.C.
Karabuk University
Institute of Graduate Programs
Department of Metallurgical and Materials Engineering
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I certify that in my opinion the thesis submitted by Hussein Basim Baqer OBAIDA titled "DEVELOPMENT AND CHARACTERIZATION OF BIOMEDICAL POROUS Ti-15Mo ALLOY" is fully adequate in scope and in quality as a thesis for the degree of Master of Science

Assoc.Prof. Dr. Hüseyin DEMİRTAŞ ....
Thesis Advisor, Department of Metallurgical and Materials Engineering

This thesis is accepted by the examining committee with a unanimous vote in the Department of Metallurgical and Materials Engineering as a Master of Science thesis. August 1, 2023

Examining Committee Members (Institutions)	Signature
Chairman: Assist. Prof. Dr. Alper İNCESU (KBU)	....
Member : Assist. Prof. Dr. Mehmet AkkAŞ (KU)	....
Member : Assoc. Prof. Dr. Hüseyin DEMİRTAŞ (KBU)	....

The degree of Master of Science by the thesis submitted is approved by the Administrative Board of the Institute of Graduate Programs, Karabuk University.

Prof. Dr. Müslüm KUZU ....
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Hussein Basim Baqer OBAIDA

ABSTRACT

Master Thesis

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Hussein Basim Baqer OBAIDA

Karabük University
Institute of Graduate Programs
The Department of Metallurgical and Materials Engineering

Thesis Advisor:
Assoc.Prof.Dr. Hüseyin DEMİRTAŞ
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Titanium alloys are widely used in biomedical applications due to their excellent biocompatibility, mechanical properties, and corrosion resistance. Porous Ti-15Mo-2In alloy is a promising material for biochemical applications due to its excellent biocompatibility, mechanical properties, and ability to promote osseointegration. In this study, prepared porous Ti-15Mo-2In alloy using a powder metallurgy technique and evaluated its suitability for biomedical applications.

The porous Ti-15Mo-2In alloy was fabricated using salt (NH_4HCO_3) as a pore-forming agent in different concentrations (20, 30, 50, and 60 Vol. %). The material's mechanical properties, including compressive strength, elastic modulus, and hardness, were characterized using standard testing methods. The compressive strength and hardness decrease with an increase in the porosity; the presence of porosity in a titanium alloy can have a negative impact on its hardness by reducing its density,

creating stress concentration points, and affecting its microstructure. Therefore, optimizing porosity during production is essential to ensure the material has the desired mechanical properties. Produced Ti alloys have been shown to have promising antibacterial properties against S. aureus and other microorganisms, which could make them a valuable material for use in medical implants and other applications where bacterial colonization is a concern. The absence of an inhibition zone for Ti alloy against E. coli could be due to various factors, including bacterial resistance, different growth requirements, or experimental conditions.

In conclusion, porous Ti-15Mo-2In alloy prepared using powder metallurgy is suitable for biochemical applications due to its excellent biocompatibility, mechanical properties, and ability to promote cell growth. Further studies are needed to evaluate its performance in vivo and to optimize its surface properties for specific biochemical applications.

Key Word : Porous Ti-15Mo alloy, Biomaterial, Powder metallurgy, Mechanical properties, Antibacterial.

Science Code : 91501

ÖZET

Yüksek Lisans Tezi

GÖZENEKLİ Ti-15Mo BİYOMEDİKAL ALAŞIMININ GELİŞTİRİLMESİ VE KARAKTERİZASYONU

Hussein Basim Baqer OBAIDA

Karabük Üniversitesi
Fen Bilimleri Enstitüsü
Metalurji ve Malzeme Mühendisliği

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Doç. Dr. Hüseyin DEMİRTAŞ
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Titanyum alaşımları, mükemmel biyouyumlulukları, mekanik özellikleri ve korozyon dirençleri nedeniyle biyomedikal uygulamalarda yaygın olarak kullanılmaktadır. Gözenekli Ti-15Mo-2In alaşımı, mükemmel biyouyumluluğu, mekanik özellikleri ve osseointegrasyonu teşvik etme yeteneği nedeniyle biyokimyasal uygulamalar için umut verici bir malzemedir. Bu çalışmada, toz metalürjisi tekniği kullanılarak gözenekli Ti-15Mo-2In alaşımı hazırlanmış ve biyomedikal uygulamalar için uygunluğu değerlendirilmiştir.

Gözenekli Ti-15Mo-2In alaşımı, %53,83 gözeneklilik ile farklı oranlarda (Hacim %20, 30, %50 ve 60 Hacim) gözenek oluşturucu madde olarak tuz (NH_4HCO_3) kullanılarak imal edilmiştir. Basınç dayanımı, elastik modül ve sertlik dahil olmak üzere malzemenin mekanik özellikleri, standart test yöntemleri kullanılarak karakterize edilmiştir. Gözenekliliğin artmasıyla bisma dayanımı ve sertlik azalır, bir titanyum alaşımında gözenekliliğin varlığı, yoğunluğunu azaltarak, stres konsantrasyon noktaları oluşturur ve mikro yapısını etkileyerek sertliği üzerinde olumsuz bir etkiye sahip olabilir. Bu nedenle, malzemenin istenen mekanik özelliklere sahip olmasını sağlamak için üretim sürecinde gözenekliliği en aza indirmek önemlidir. Ti alaşımlarının S. aureus ve diğer mikroorganizmalara karşı ümit verici antibakteriyel özelliklere sahip olduğu gösterilmiştir, bu da onları tıbbi implantlarda ve bakteriyel kolonizasyonun önemli olduğu diğer uygulamalarda kullanım için değerli bir malzeme haline getirebilir. Ti alaşımı için E. coli'ye karşı bir inhibisyon bölgesinin olmaması, bakteri direnci, farklı büyüme gereksinimleri veya deneysel koşullar dahil olmak üzere çeşitli faktörlere bağlı olabilir.

Sonuç olarak, toz metalürjisi kullanılarak hazırlanan gözenekli Ti-15Mo-2In alaşımı, mükemmel biyouyumluluğu, mekanik özellikleri ve hücre büyümesini teşvik etme yeteneği nedeniyle biyokimyasal uygulamalar için uygun bir malzemedir. Performansını in vivo olarak değerlendirmek ve belirli biyokimyasal uygulamalar için yüzey özelliklerini optimize etmek için daha ileri çalışmalara ihtiyaç vardır.

Anahtar Sözcükler : Gözenekli Ti-15Mo alaşımı, Biyomateryal, Toz metalurjisi, Mekanik özellikler, Antibakteriyel.

Bilim Kodu : 91501

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SYMBOLS

α : Alpha

β : Beta

P : Density

E : Young’s modulus

μm : Micrometer

°C : Degrees Celsius

ABBREVIATIONS

PM : Powder metallurgy

ASTM : American Society for Testing and Materials

JCPDS : Joint Committee for Powder Diffraction Standards

Ra : Average Roughness

EDS : Energy-Dispersive Spectrometry

CP-Ti : Commercial grade titanium

PART 1

INTRODUCTION

1.1 INTRODUCTION

Biomaterials Science, Tissue Engineering, Materials Science, Biomedical and Medicine fields are critically important for countries, and their contribution to social welfare is indisputable. In addition, these fields play a leading role in the development of the science, technology, and health sector of the countries. The expenditures made by the countries in these areas and the share allocated to these expenditures from the state budgets are at the highest level. Biomaterials are primarily used in medical applications, but they are also used in biotechnology. Biomaterials have an important place in the field of medicine. Local manufacturers in the medical sector can operate in a limited and specific product device range due to low technological manufacturing, R&D structures and medical activity approval (American Food and Drug Administration approval, USA Food and Drug Administration, FDA) certificate deficiencies [1] .

Today, the number of patients in orthopedic surgery increasing every year accelerates the progress of the orthopedic market. This leads to a steady increase in demand with the number of surgical operations and rapid recovery after surgery. Loss of implant stability, infection, abrasion, periprosthetic fracture, tissue incompatibility and aseptic losses can be counted among the problems frequently encountered in patient complaints after orthopedic applications [2, 3]. Although titanium and its alloys are the most widely used metallic materials in orthopedic devices, they are preferred because of their high biocompatibility with bone tissue, high corrosion resistance, high strength and modulus of elasticity [4]. Pure titanium and Ti6Al4V alloy are frequently used in biomedical applications. Generally, pure titanium with a tensile strength of 240-740 MPa is used in dental implants. The reason why Ti6Al4V alloy is highly preferred is mainly due to its high corrosion resistance, low density and high static and dynamic strength.

Metallic implants have higher strength and rigidity than bone [5]. However, Ti6Al4V is harder than bone tissue. For this reason, studies are carried out in the United States for the development of titanium alloy materials with less hardness [6]. Where the implant is used instead of hard tissue, the evaluation of biocompatibility and biofunction is made with at least three criteria. These three criteria are: (1) mechanical compatibility (2) morphological compatibility (3) biological compatibility (biocompatibility) [7].

In the last decade, it is known that most of the personalized hip and knee implants have been produced from Ti6Al4V material with various manufacturing techniques [8]. Among the production techniques used, selective laser melting method (SLM), electron beam melting (EBM) and rapid production (RM) seem to have a special place [9, 10]. In the literature researches, it was also encountered with studies that the microstructure and mechanical behavior properties of the parts produced from Ti6Al4V alloy material by forging, which is the thermomechanical method, electron beam melting (EBM) and selective laser melting methods were compared . In these studies, besides the microstructure phase morphology, the effect of deformation and martensite substructure on mechanical behavior, especially tensile strength, was emphasized [10]. As a result of the studies, it has been understood that relatively simple geometry implants can be made by EBM and SLM methods [10]. It has been determined that implants made of Ti6Al4V alloy material by EBM and SLM methods have similar mechanical behaviors and microstructures to forged or cast ones. The tensile strength of implants produced by EBM and SLM method varies between 1 and 1.45 GPa [10-13]. The place of use of orthopedic materials in the body should be determined by taking into account the weight and daily activities of the person. In the selection of implant materials; Mechanical loads are at the forefront and the selection of materials with mechanical properties that will provide strength is very important. Although the modulus of elasticity of metallic biomaterials is very high (200GPa in 316L stainless steel, 110GPa in titanium), this value is 10-15 GPa in human bone [1]. It is known that alloys with a modulus of elasticity closer to human bone carry less stress.

As a result of scientific studies, it has been found that the upper and lower limit values of the average roughness value (1-50μm) and the average particle size (10-500μm) for the surface morphology are suitable for a successful implant, regardless of the type of implant material (metallic, ceramic, polymeric) [14]. If the particle size is less than 10 μm, it can cause too much poisoning for the surface fibroplastic cells, and the physical presence of this poison has a negative effect on the cells. If the pores are larger than 500 μm, the structural integrity cannot be maintained as the surface will be too rough. This is called “morphological compatibility” [14]. The methods used in titanium implant surface modification are morphological methods that aim to increase cell adhesion, grouped as physicochemical, morphological or biochemical. Bagno and DiBello [15] stated that cell functions and biochemistry can be understood with appropriate surface modification and examined the modifications under three main headings: mechanical, chemical and biochemical methods. Mechanical methods: These are the methods that shape the surface with physical forces. The most frequently used mechanical techniques are; turning, cutting, sanding.Chemical methods: applied to change the chemical structure of titanium, especially the surface layer [16]. It consists of the reactions that take place between the titanium surface and the chemical solution used. Acid, alkali, NaOH-H_2O_2, H_2O_2 or heat reactions are the main ones [16, 17]. Etching and roughening of titanium surfaces using strong acids such as hydrofluoric acid (HF), hydrochloric acid (HC1), nitric acid (HNO_3), sulfuric acid (H_2SO_4) is a widely used method. It has been reported that 1.5 - 2 mm diameter micro pits on the implant surface are formed by pickling [18].

Some alloys based on titanium still have insignificant participation in the alloy market for applications such as biomaterials. The reasons for this are diverse and include the high cost and lack of data for application projects for these alloys. According to Li et al.[19], there is little literature on porous titanium-molybdenum with indium alloy to fix the stress shielding of Ti-15Mo alloy by adding Indium to the base alloy. The porosity range, between 35 and 40%, and the small pore size of the porous Ti-Mo alloy prepared by Kolobov et al.[20]. It is imperative to manufacture porous titanium-molybdenum alloys with wide ranges of porosity, pore size, and mechanical properties. The main objective is to get suitable stress shielding of Ti-15Mo alloy by adding Indium to the base alloy. The following steps can summarize this aim:

- Processing and characterizing of porous Ti-15Mo alloy via powder metallurgy to obtain materials suitable for use as bioimplants.
- Study the microstructure changes, physical and mechanical properties as a function of the addition of Indium and pore-forming agent.
- Investigating the the antibacterial characteristics of porous Ti-Mo-In alloy.

PART 2

LITERATURE REVIEW

Due to their high strength-to-weight ratio, low Young's modulus, and biological compatibility, titanium alloys are promising metallic materials for long-term bone implant applications. Researchers focused on modifying the surfaces of these alloys to increase wear resistance and bioactive properties because of their low corrosion and abrasion resistance.

In his work, the motivation for the study was two parts. The first motivation concerns the preparation of porous Ti-15Mo-2In alloys using different concentrations of salt as pore-forming agents. The second motivation was the investigation of physical, Antibacterial, and mechanical properties, including compressive strength, hardness, and Young's modulus.

Ti-15Mo alloy is a biocompatible material that has been widely studied for its potential use in biomedical applications, such as orthopedic and dental implants. Here is a brief literature review of some of the key studies on Ti-15Mo alloy:

"Microstructure and mechanical properties of Ti-Mo alloys for biomedical applications" [21] . This study investigated the microstructure and mechanical properties of Ti-15Mo and Ti-20Mo alloys, and compared them to Ti-6Al-4V alloy. The results showed that Ti-15Mo and Ti-20Mo alloys had lower strength and higher ductility than Ti-6Al-4V alloy, making them suitable for biomedical applications.

" Mechanical properties, in vitro corrosion resistance and biocompatibility of metal injection molded Ti-12Mo alloy for dental applications " [22]. This study evaluated the corrosion behavior and cytotoxicity of Ti-15Mo and Ti-20Mo alloys in simulated body fluid and human osteoblast cells. The results showed that Ti-15Mo and Ti-20Mo alloys had good corrosion resistance and no cytotoxicity, indicating their potential use in biomedical applications.

"In vivo biocompatibility and corrosion behavior of Ti-15Mo alloy for dental implant application" [23].This study investigated the in vivo biocompatibility and corrosion behavior of Ti-15Mo alloy in a rabbit model for dental implant application. The results showed that Ti-15Mo alloy had good biocompatibility and low corrosion rate, suggesting its suitability for dental implant applications.

"Effect of Mo content on the mechanical properties and biocompatibility of Ti-Mo alloys" [24] .This study investigated the effect of Mo content on the mechanical properties and biocompatibility of Ti-Mo alloys with Mo content ranging from 5 wt% to 30 wt%. The results showed that Ti-15Mo alloy had the best combination of mechanical properties and biocompatibility among the Ti-Mo alloys tested.

Elastic softening of β-type Ti–Nb alloys by indium (In) additions. " [25]. This study investigated the effect of indium (In) additions on the elastic properties of β-type Ti-Nb alloys. The authors prepared Ti-Nb alloys with varying amounts of In (0-5 wt%) using arc melting and investigated their microstructure and elastic properties using various techniques, such as X-ray diffraction, scanning electron microscopy, and resonant ultrasound spectroscopy. The results showed that the addition of In led to a decrease in the elastic modulus and an increase in the Poisson's ratio of the alloys. The authors attributed this to the formation of a β-phase with a higher Nb content and a more disordered structure due to the addition of In. The study suggests that the addition of In can be used to tailor the elastic properties of Ti-Nb alloys for specific applications, such as implant materials.

"Effect of annealing temperature on microstructure and mechanical properties of Ti-15Mo-2In alloy"[26]. This study investigated the effect of annealing temperature on the microstructure and mechanical properties of Ti-15Mo-2In alloy. The results showed that the alloy had good strength and ductility after annealing at 750°C, which is suitable for biomedical applications.

"Effects of Mo contents on the microstructure, properties and cytocompatibility of the microwave sintered porous Ti-Mo alloys"[24]. This study investigated the effects of molybdenum (Mo) content on the microstructure, properties, and cytocompatibility of porous Ti-Mo alloys prepared by microwave sintering. The authors prepared Ti-Mo alloys with Mo contents ranging from 5 wt% to 25 wt% and evaluated their microstructure, mechanical properties, and cytocompatibility using various techniques, such as scanning electron microscopy, X-ray diffraction, and compression testing. The results showed that the addition of Mo led to a decrease in the grain size and an increase in the porosity of the alloys. The mechanical properties of the alloys, such as compressive strength and elastic modulus, were also improved with the addition of Mo. The cytocompatibility of the alloys was evaluated using human osteoblast cells, and the results showed that the alloys with 10-20 wt% Mo had good cytocompatibility. The study suggests that the addition of Mo can improve the microstructure, mechanical properties, and cytocompatibility of Ti-Mo alloys, and that microwave sintering can be an effective method for preparing porous Ti-Mo alloys with tailored properties for biomedical applications.

"Corrosion Behavior of Ti-15Mo Alloy in Simulated Body Fluid"[27] . This study evaluated the corrosion behavior of Ti-15Mo-2In alloy in simulated body fluid. The results showed that the alloy had good corrosion resistance, indicating its potential use in biomedical applications.

"Investigation of the effect of indium addition on the mechanical and electrochemical properties of the Ti–15Mo biomedical alloy" [28].

The objective of this study was to investigate the effect of indium addition on the mechanical and electrochemical properties of Ti-15Mo alloy for biomedical applications. The results showed that the addition of indium to Ti-15Mo alloy improved its mechanical properties, such as hardness and tensile strength. The alloy with 2 wt% indium had the highest hardness and tensile strength, which was attributed to the formation of a fine-grained microstructure. However, the addition of more than 2 wt% indium led to a decrease in mechanical properties due to the formation of a coarse-grained microstructure. The electrochemical behavior of the alloys was also improved with the addition of indium, as evidenced by a decrease in the corrosion rate and an increase in the polarization resistance.

Overall, these studies suggest that Ti-15Mo-2In alloy is a promising material for biomedical applications due to its good mechanical properties, corrosion resistance, biocompatibility, and antibacterial properties. Further studies are needed to explore its potential use in specific applications and to optimize the porosity for different biomedical applications.

PART 3

THEORETICAL BACKGROUND

3.1 BIOMATERIALS

A biomaterial is a pharmacologically inert material designed to be implanted inside a living organism [5]. The biomaterial is implanted to replace or regenerate tissues that have suffered damage or been lost due to trauma, malformations or degenerative diseases. While for most of the twentieth century, industrial materials that were as inert as possible were used to manufacture biomedical devices, currently, biomaterials for medical use are specifically designed and processed for this purpose. In the design of biomaterials, not only the mechanical, chemical and physical properties required in materials for industrial use are taken into account, but also biological requirements such as biocompatibility, bioactivity and osseoinduction. Biomaterials must meet three basic requirements: be biocompatible, resist corrosion from body fluids, and have good mechanical properties to withstand the applied physiological loads. The transcendental property in biomaterial human body interaction is biocompatibility, the absence of pernicious physicochemical reaction of the implanted biomaterial with tissues and biological body fluids. The lack of biocompatibility induces an adverse reaction between the implant and its biological environment, progressively creating irritation, inflammation or infection to such a degree that the implant must be removed to avoid tissue destruction or impaired organ function, nearby with severe consequences for the health of the patient [29]. According to their chemical composition, biomaterials are classified as metallic, polymeric, ceramic, and composite, according to their origin, natural and synthetic, and by their structure, compact and porous. In manufacturing biomedical devices adapted to the human body, the four mentioned engineering materials are used depending on whether it is a hard or soft tissue to be replaced or repaired. In bone replacement surgeries, such as hip or knee arthroplasties and for orthopedic fixations in osteosynthesis, mainly metallic materials are used. Ceramic and polymeric materials are restricted due to their intrinsic brittleness or low mechanical resistance.

Biomaterials are compatible with tissues in the body. Since they are in this structure, they do not harm the body. Our body generally does not accept foreign elements added to its structure from the outside. Therefore, materials added to the body are initially perceived as a threat to the body. At this stage, the concept of biocompatibility is needed [30].

Biocompatibility can be defined as the materials added to our body that are not considered an external threat to the body. These materials, which are not accepted externally, adapt to the body and perform their functions. When biomaterials are placed in the body, many other reactions occur in addition to their normal reactions. The main reactions that may occur in the body, Interaction between proteins and biomaterials in our tissues, can be shown as increased red blood cell count, tumour production and immune system activity [31].

Biomaterials react with tissues when placed in the body. These reactions are divided into toxic (toxic) and non-toxic (non-toxic) reactions. If the reaction is toxic, the living tissues do not accept the substance, and the surrounding tissues die. Biomaterials are considered bioactive or bioinert if the answer is non-toxic. Material; If it is considered bioinert, it causes fibrous tissue of varying thickness to form on the implant material. Material; If it is bioactive, it causes a strong bond between the material and the tissue. Another possibility is that the material is resorbable and absorbed by and replaces the surrounding tissues. The reactions of tissues to external (foreign) substances are given in Table 3.1.

Table 3.1. Reactions of tissues to external substances [32].

Response Given	Tissue Response
Toxic	Surrounding tissues die
Non-toxic/Biologically non-active	Produces fibre-like tissue of variable thickness
Non-toxic/Biologically active	Formation of fibre-like tissue of variable thickness
Non-toxic/ Absorbable	Surrounding tissue replaces material.

The body's responses to biomaterials have been very different. Under certain conditions, some materials were accepted by the body, while the same materials could be rejected by the body when conditions changed. In the last three decades, important information has been gained in understanding biomaterial and tissue interactions.

Biomaterials are used in the human body environment, which has very variable conditions. For example, the pH level in body fluids varies between 1 and 9 according to different tissues. It is known that during our daily activities, our bones are subjected to a stress of approximately 4 MPa and our tendons 40-80 MPa. The average load on any hip joint can be up to 3 times our body weight and up to 10 times our body weight during activities such as jumping. Such body tensions are constantly repeated during standing, sitting, and running. Biomaterials should be able to withstand all these harsh conditions [33].

Biomaterials in orthopedic applications, joint prosthesis and bone prosthesis materials, facial and maxillofacial surgery, dental implants, artificial heart parts, heart valves, catheters, fixation materials, spinal instrumentation, metal parts, screws, eye screws, screw washers, nails, fixation wires, hip plates, anatomical plates, angled plates and occasionally implantable devices, etc. [34].

The most commonly used metallic biomaterials are stainless steels (316L), titanium and titanium alloys, cobalt-chromium alloys, cobalt-nickel-chromium-molybdenum alloys, tantalum alloys, nickel-titanium alloys, amalgam and gold. Due to the low mechanical strength of elements such as platinum, tantalum and zirconia, their use as implant materials is limited. The metal materials most commonly used as load bearings are stainless steel, Co-Cr-Mo alloys or titanium and titanium alloys [35]. In Figure 3.1, the biomaterials used in the human body are shown schematically.

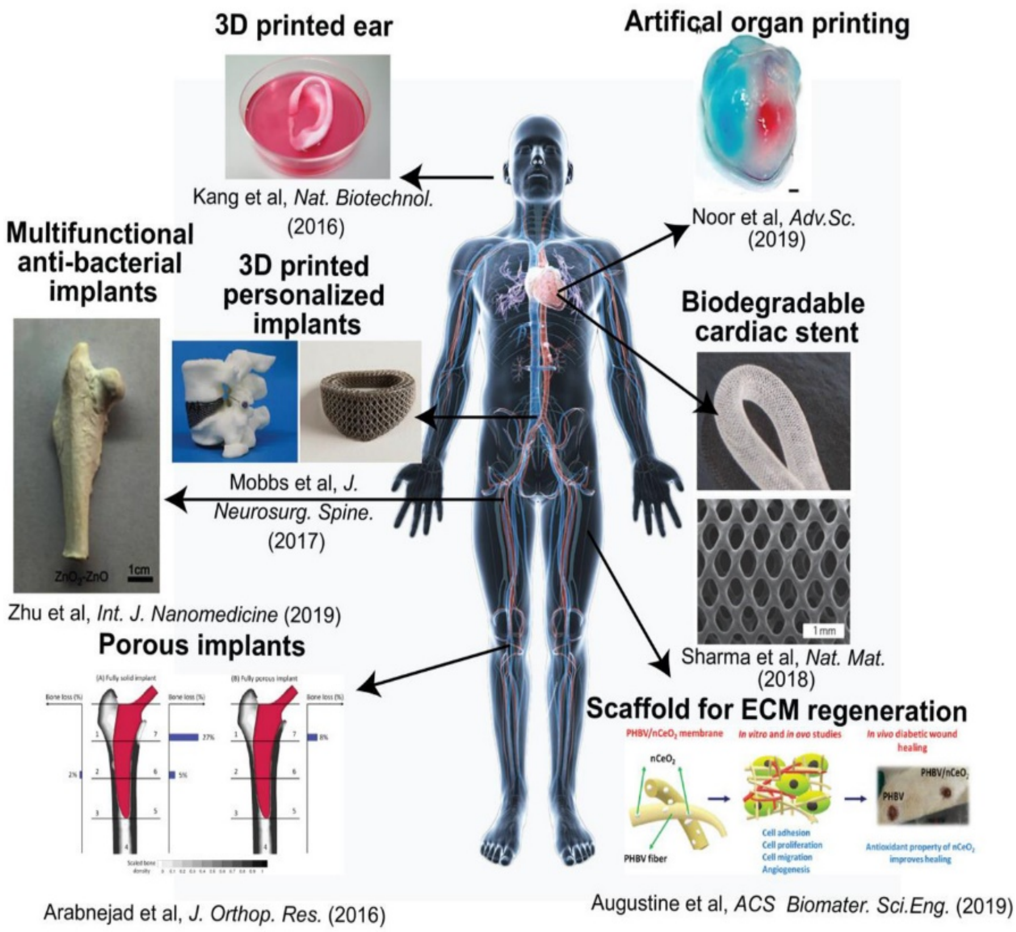


Figure 3.1 The use of biomaterials in the body [35]. [35][35][35]

3.2 METALLIC BIOMATERIALS

Metallic biomaterials are the most suitable materials due to their resistance to the mechanical conditions of the body's musculoskeletal system. Metallic biomaterials are preferred because they can withstand heavy, long-term, variable or sudden loads without losing their properties within predetermined limits [29].

Metallic biomaterials are classified according to pure metal or alloy elements [36]. These;

- Stainless steel
- Cobalt-Chromium (CoCr) alloys
- Titanium (Ti) alloys.

As a result of the experimental measurement that the femoral head is loaded at the rate of 3.5 times our body weight (usually 80 kg body weight) during walking, it is seen that total hip prostheses must be sufficiently resistant to these loads that can be determined. In addition, these prostheses must be resistant to abrasion caused by friction in the joint.

Today, as a hip replacement material, vitalium (Co-Cr-Mo alloy), stainless steel, high-density polyethylene, polymethylmethacrylate and Al_2O_3 type ceramics are used [37]. Due to the low mechanical strength of elements such as platinum, zircon or tantalum, their use as implants is limited. As stated before, frequently used as load-bearing metallic biomaterials; series of stainless steels (316L), cobalt-chromium-molybdenum (Co-Cr-Mo) alloys, titanium and titanium alloys [38].

Pure titanium or Ti6Al4V alloy is commonly used in biomedical applications. Generally, pure titanium with a tensile strength of 240-740MPa is used in dental implants. On the other hand, Ti6Al4V alloy material is the most common titanium alloy. The usage rate in the world titanium market reaches 50%. Such a preference for Ti6Al4V alloy material is due to its high static and dynamic strength, high corrosion resistance and low density. Metallic implants have higher durability and modulus of elasticity than bone [39]. A comparison of the properties of metallic biomaterials as implants is given in Table 3.2, and 15 implant applications of metallic biomaterials are given in Table 3.3. In addition, the hardness values of metallic biomaterials and cortical bone are shown graphically in Figure 3.2 [40].

Table 3.2. Comparison of the properties of metallic biomaterials as implants [40, 41].

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

Table 3.3. Implant applications of metallic biomaterials [40].

Materials	Application
Ti and it's alloys	Bone and joint replacement, fracture fixation, dental implants, pacemaker, encapsulation
Co-Cr alloy	Bone and joint replacement, dental implants, dental restorations, heart valves
Stainless steel	Fracture fixation, stents, surgical instruments
Ni-Ti alloy	Bone plates, stents, orthodontic wires
Platinum and Pt-Ir	Electrodes
Hg-Ag-Sn (amalgam)	Dental restorations

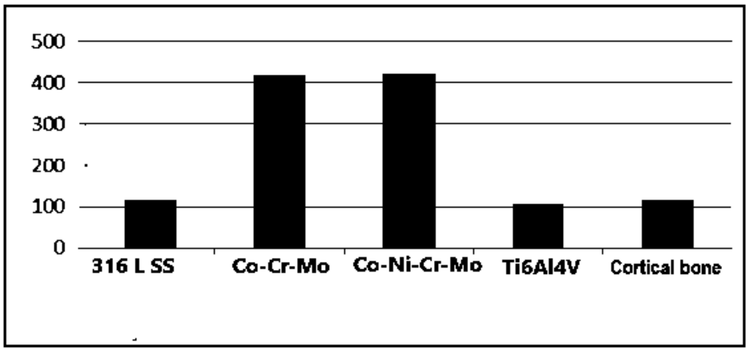


Figure 3.2. Hardness values of metallic biomaterials and cortical bone [42].

Although the modulus of elasticity of metallic biomaterials is very high (200 GPa in 316L stainless steel, 110 GPa in titanium), this value is 10-15 GPa in human bone. This mechanical incompatibility causes the implants to be structurally harder than human bone tissue. Alloys with modulus of elasticity closer to that of human bone are subject to relatively less stress. The hardness of metallic biomaterials is related to the modulus of elasticity [43].

Since the modulus of elasticity of stainless steel is higher than that of titanium, it has a higher hardness than titanium. Regarding strength and elasticity, titanium alloys are significant in metallic biomaterials. Although stainless steel materials show less tensile strength and fatigue strength, their ductility is high. Pure titanium, tantalum, and niobium have low fatigue strength and high elongation at break. Figure 3.3 shows the elastic modulus of different alloys used as implants [1].

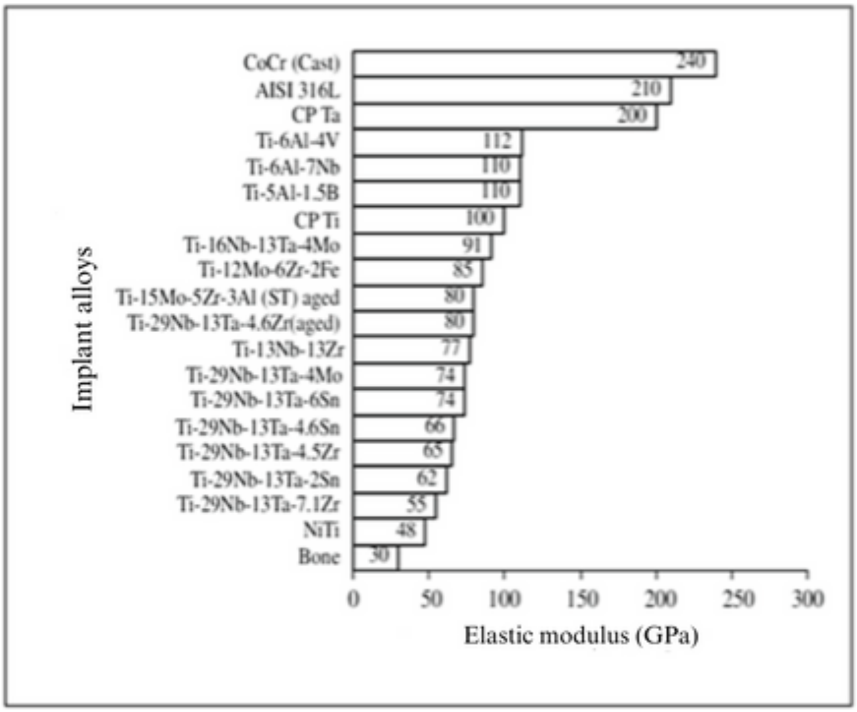


Figure 3.3 Elastic modulus of different alloys used as implants [1][1][1][1].

Due to their crystalline structure and strong metallic bonds, metal and metal alloys with superior mechanical properties have a high share in biomaterials. The joint prosthesis and bone regeneration material used in orthopedic applications are metal alloy, and they are also used as facial implant materials and cardiovascular surgery, in facial, maxillofacial surgery, artificial heart parts, catheters, valves and heart valves. The first metal developed for use in the human body is known as "Sherman-Vanadium Steel." Many metals used in producing biomaterials, such as iron, copper, chromium, cobalt, nickel, titanium, tantalum, molybdenum and vanadium, are suitable for applications in the living body, provided they are used in small quantities. These metals, known to be harmful to the body, can also occur during metabolic activities. Examples are cobalt synthesis from vitamin B12 or iron formation as a cellular function [44]. Today, three main metal groups and their derivatives are used as prosthetic materials in orthopedic surgeries. These materials are; stainless steel materials, cobalt-chromium alloys, and titanium alloys. The most widely used pure or alloyed biometals are; Co-Cr-Mo, Co-Cr- W-Ni, Ti-Al-V, Co-Ni-Cr-, Mo-Ti alloys and pure titanium and stainless steel types [45]. Besides noble metals, metals are found in the earth's crust in mineral form. They are chemically combined with other metals as; an example is a metal oxide. These metals must be found, removed and separated by pure metal transition processes to make them more suitable. The raw metal products obtained are supplied to the producers as ingots. The untreated metal products are further pre-treated in metal implant alloys containing more than one component. Examples include processes such as remelting, the addition of alloying elements and solidification. After such steps, alloys with the desired specific chemical properties are obtained.

Metallic biomaterials are classified according to metal and alloying elements. The most commonly used pure and alloys as metallic biomaterials; are alloys of Ti, Cr, Co, W, V, Al, Ni, Mo and stainless steels. Metals are rarely used alone because of their insufficient properties. However, the improvement of its properties with the addition of alloying elements has made their use widespread and convenient. Stainless steels, titanium-based alloys and cobalt-chromium alloys are metal groups used in orthopedic surgery. The most used in metallic biomaterials, Ti-Al-V alloys are pure titanium and stainless steels with Co-Cr-W-Ni, Co-Cr-Mo, and Co-Ni-Cr-Mo-Ti.

It is vanadium steel, used in manufacturing Sherman plates and screws, which are used in alloy fracture treatments first in the human body. Implant alloys used in the body (metal alloys such as chromium, cobalt, iron, nickel, tantalum, niobium, titanium, tungsten and molybdenum) cannot be used safely in the body for a long time due to their corrosion and toxic effects [46].

In the 19th century, steel materials began to be used as plates or screws to treat bone fractures. Treatment of the bone with screws provided a stronger recovery compared to the previously used wire repair technique. Vanadium and nickel steels have replaced carbon steel with low corrosion resistance. However, these materials are not one hundred percent resistant to corrosion. As a result, it showed toxic and allergic effects in the body. As a result, the basic biomedical materials used in orthopedic applications; cobalt-chromium-molybdenum alloys (vitalium), titanium and titanium alloys, and stainless steels [47].

It is desired that the fatigue strength of the implants used in the human body is good. We can understand the importance of this, especially if an implant is to be used in hip connections. Assuming that a healthy person takes an average of one million steps per year, it is predicted that this person's hip connections will be weighed two or three times his body. This situation has prevented the use of cast stainless steel in orthopedics due to its low fatigue strength and coarse grain size. The most useful in stainless steel is low carbon 316L. However, it should not be overlooked that stainless steel cannot be used in all areas due to the poor compatibility between texture and metal [48].

Cobalt-chromium is the most widely used cobalt alloy for biomaterial purposes. It has high elastic modulus, hardness and strength values and high corrosion resistance compared to stainless steel. It has weaker properties in terms of formability and ductility. The most commonly used are cobalt-chromium-molybdenum and cobalt-nickel-molybdenum alloys. Cobalt-chromium-molybdenum in artificial joint and dental applications; Cobalt-nickel-molybdenum is used more in the hip, knee and joint prostheses working under load.

Titanium and its alloys started to be used in biomaterials towards the end of the 1930s due to their lightness compared to stainless steel and cobalt alloys. Its superior biocompatibility and non-toxicity cause titanium to be among the most widely used metals [49].

3.3 TITANIUM AND ITS ALLOYS

3.3.1. Titanium

Titanium is the ninth most abundant element in the Earth's crust, with 0.8% by weight, although it does not frequently exist in concentrations that make its extraction economically viable. It is mainly present in igneous rocks, the sediments derived from them, and many silicates replacing silicon. Metallic titanium, in the state in which it is mined from ore, is called sponge titanium. This is due to its porous aspect that gives the appearance of a sponge. In the ore from which it is extracted, it is found to form rutile (TiO_2) or ilmenite ($FeTiO_3$) [50].

Titanium is classified as a light metal, with a density of 4.507 g/cm^3, half that of iron and twice that of aluminum. They provide high specific resistance, a high melting point, excellent corrosion resistance and excellent biocompatibility, although they have poor thermal and electrical conductivity. It also stands out that its cost is high. The basic characteristics of titanium and its alloys are shown in Figure 3.4 compared to other metallic materials: iron, nickel, and aluminum. Although titanium has the best specific properties (relationship between resistance and density), it is intended only for specific applications in certain areas due to its high cost. This cost is mainly due to its significant reactivity with oxygen, making it necessary to work under vacuum or inert atmosphere conditions to obtain and process. On the other hand, its significant reactivity with oxygen allows the immediate formation of a TiO_2 oxide layer when exposed to air. This layer does not come off easily; it adheres very well (as in aluminum) and gives it excellent resistance to corrosion and contamination in different aggressive media, especially in aqueous acid media.

	Ti	Fe	Ni	Al
Melting Temperature (°C)	1670	1538	1455	660
Allotropic Transformation (°C)	$\beta \xrightarrow{882} \alpha$	$\gamma \xrightarrow{912} \alpha$	-	-
Crystal Structure	bcc → hex	fcc → bcc	fcc	fcc
Room Temperature E (GPa)	115	215	200	72
Yield Stress Level (MPa)	1000	1000	1000	500
Density (g/cm^3)	4.5	7.9	8.9	2.7
Comparative Corrosion Resistance	Very High	Low	Medium	High
Comparative Reactivity with Oxygen	Very High	Low	Low	High
Comparative Price of Metal	Very High	Low	High	Medium

Figure 3.4 Most important characteristics of titanium and the most commonly used metals: iron, nickel and aluminum [51].

The melting temperature of titanium (1668 ºC) is much higher than that of aluminum, its competitor in applications requiring lightness. This is an advantage in applications where the temperature exceeds 150 ºC. The high reactivity of titanium with oxygen limits its use at maximum temperatures of approximately 600ºC, since at higher temperatures, the diffusion of oxygen through the protective oxide layer is very rapid, which causes excessive growth of the oxide layer and embrittlement of adjacent layers, very rich in oxygen [51].

The main barrier to titanium alloys in commercial applications is the cost of semi-finished solid goods. This high cost includes the need for complex operations during extraction and separation, lack of sufficient resources to cover plant investments and high transaction costs. Titanium alloys stand out with two different properties. Their high strength and superior corrosion resistance have made these metals and their alloys indispensable for aerospace, chemical, and medical engineering. Despite these positive properties, titanium is inadequate in mobile contact engineering applications due to its frictional wear properties. Titanium's weak tribological properties, and high and variable friction coefficient, limit the usage areas of titanium and its alloys. Therefore, its alloys with titanium are generally chosen for applications where wear is not risky. Energy and material loss occur due to friction and wear in applications where titanium is used. As a result of wear, the protective oxide layer, which creates corrosion resistance, is deteriorated.

For this reason, besides wear, dangerous corrosion occurs. In order to improve the friction wear properties of titanium, it seems essential to apply surface modification techniques. In order to improve these weak features, various surface treatment applications are made today, and very successful results can be obtained [52, 53]. Today, over one hundred titanium alloys are known. Only 20 to 30 of these can be used in commercial applications. Among conventional alloys, Ti6A14V alloy material alone makes up 50% of the total titanium alloy used. On the other hand, unalloyed titanium accounts for 20% to 30% of the entire amount used [54].

3.3.2 Crystal Structure

Titanium can crystallize in more than one structure. These structures are stable as a function of temperature. Pure titanium exhibits an allotropic phase transformation at 882°C, changing from a body-centred cubic crystal structure (α phase) above the transformation temperature to a close-packed hexagonal structure (β phase) below said temperature. The exact temperature at which the transformation occurs depends strongly on this metal's interstitial and substitutional elements. It depends, ultimately, on the purity of the material. Figure 3.5 shows the unit cells of both structures with the value of the network parameters corresponding to room temperature [51].

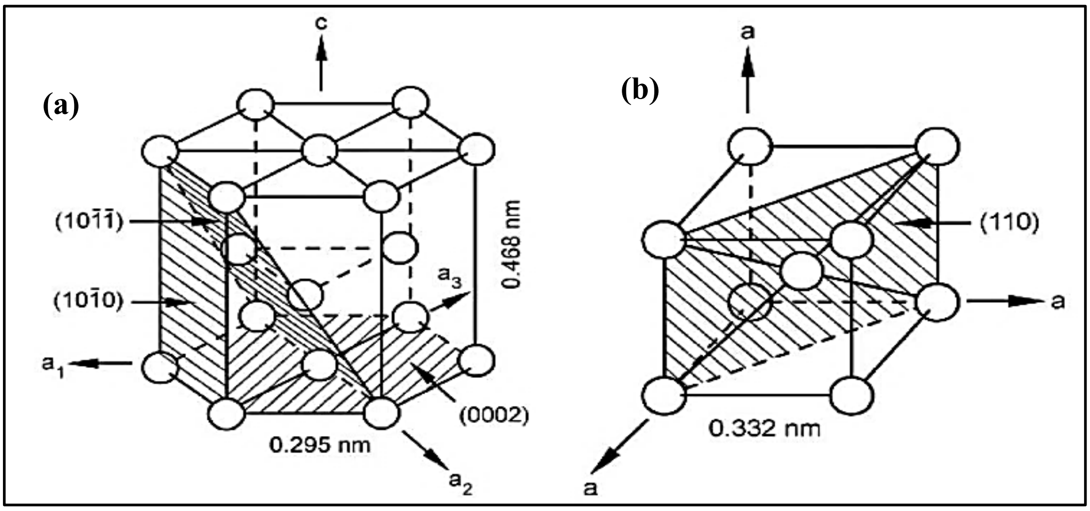


Figure 3.5. The crystalline structure of titanium: (a) α-titanium: compact hexagonal structure, (b) β-titanium: body-centred cubic structure [51].

The existence of two different crystalline structures makes it possible to carry out thermal treatments with total transformation since the allotropic forms present a different behavior when faced with deformation:

- Phase α is deformable and has superior to corrosion resistant
- β phase, easily deformable.

The exact temperature at which the transformation occurs depends strongly on the interstitial and substitutional elements found in the titanium. Therefore it depends on the purity of the material. Alloying elements that change the allotropic transformation temperature can be divided into four groups, α-stabilizers (Al, O, N, C), β-stabilizers (which is divided into 2: β-isomorphs (V, Mo, Nb, Ta) and β-eutectoid (Fe, Mn, Cr, Ni, Cu, Si)) and neutralizers (Zr, Sn). Certain additions, such as tin, produce solid solution hardening without affecting the transformation temperature [51].

3.3.3 Classification of titanium alloys

It is common to divide titanium alloys into three groups, depending on the phases present:

- α and quasi-α alloys
- α-β alloys

- β-alloys

α and quasi-α have a compact hexagonal structure at low temperatures. It usually contains aluminum, tin and/or zirconium. Preferably chosen for applications at elevated and cryogenic temperatures. A combination of cold work and cooling is used to modify their properties since they are not sensitive to heat treatments. Alloys with this type of structure are generally more resistant to hot flow than those with an α-β or β structure. Later studies revealed that adding silicon precipitated dislocation at high temperatures, preventing it from climbing and reducing its deformation ability. After that, 0.5% Si was added to the high-temperature alloys produced. Today, the most advanced high-temperature titanium alloy, American TIMETAL 834, works for a long time at 6000C and protects against oxidation [55].

In α-β contain both α and β phase stabilizing elements. These alloys retain, at room temperature, after heat treatment, more β phase than almost α-alloys. α-β alloys can improve their mechanical characteristics by solution heat treatments and aging. This treatment can increase the tensile strength between 30 and 50% compared to the resistance in the annealed state. In this group of alloys is the well-known Ti6Al4V, the most widely used titanium alloy of all those existing on the market. Approximately 45% of the titanium production is dedicated to said alloy, while 30% is destined to grades unalloyed titanium and the other 25% is divided among all the other alloys. Other alloys used after α+β alloys Ti6Al4V are; Ti-6-2-4-6 with high strength and high toughness, Ti-6-2-2-2-2, Ti-55-24-S, developed for use in gas turbine engines at temperatures up to 4000°C, high strength They are Ti-6-6-2 and IMI 550 and Ti17 alloys [56].

β-titanium alloys have a higher content of β-phase stabilizing elements and a lower α-alpha gen content than α-β alloys. Characterized by their high hardening capacity, for example, they can be hardened in air and completely retain the β phase in small thicknesses. They are titanium alloys with a better aptitude for forming by plastic deformation, able to deform in the cold much better than α or α-β alloys. Their modulus of elasticity is usually low, which is why they are suitable for some specific applications in the biomedical or aerospace field [57].

β titanium alloys contain significant amounts of β phase stabilizing alloying elements in their chemical composition. It is distinguished from other titanium alloys by its hardenability, malleability, cold forming and high-density properties. Although β titanium alloys have the same strength values as α + β alloys at room temperature, these strength values come after α + β alloys at high temperatures [58].

3.4 BIOCOMPATIBILITY OF TITANIUM ALLOYS

Titanium alloys were used as implantation materials in the 1950s but were limited to CP-Ti and Ti6Al4V alloys. Both of these alloys were produced primarily for military and industrial applications, but they have been widely used as implant materials in the medical field because they meet the desired properties [59]. Titanium and its alloys, used in dentistry and orthopedic fields, are used appropriately in the production of implants and prostheses. Researchers have preferred titanium among the metals used in the human body in recent years due to its superior properties [60].

Superior biocompatibility for insertion into the skin in the body

- Low probability of reaction
- Suitable for Magnetic Resonance as it is paramagnetic
- Light and low density compared to most metals
- Not allergic or less allergic
- Good producibility (small size)
- Good mechanical properties
- Elastic module close to bone.

Corrosion is the deterioration of the structure due to chemical or electrochemical events from the surface layer to the inner regions. Under normal conditions, liquids with a pH of 7.4 and NaCl of 0.9% are present in the human body. The most suitable metals that show resistance to corrosion in these conditions in the human body are Ti, Nb, Ta and 316L. Although corrosion resistance is good in these metals, allergic reactions and toxicity can be seen in the tissue area due to the effect of ions [61].

The oxide layer on the titanium's surface increases its corrosion resistance and provides biocompatibility. Titanium collects proteins inside the body, and many studies have been done on this subject. The oxidized surface of titanium and its alloy, which has a structure suitable for cell growth, also allows osseointegration. The oxide layer holds the H+ ion and reacts with the OH group to form TiOH. This hydroxide forms a deposition of phosphorus and calcium with apatite and combines with bone tissue. The surface of the implant material used in dentistry is usually coated with oxide. Plates and screws, first used by Jergensen in 1951 in fracture bone bonding, were produced from pure titanium. Dr. Jergensen has seen that the material has good resistance to corrosion and compatibility with tissue, but its strength has some problems. 1959 Ti6Al4V alloy was developed in the Sivash hip prosthesis center in the USSR. In the 1970s, due to its good strength values, Ti6Al4V began to replace pure titanium in the production of implants, plates, nails and screws in England. In the USA, towards the end of the 1970s, the properties of the Ti6Al4V alloy were developed, and the implant to be used in the hip was produced. In the 1980s, two different types of titanium alloys began to be used. Ti5Al2.5Fe and Ti6Al7Nb alloys are similar to Ti6Al4V alloys but differ in the absence of vanadium. Howmedica R&D established a program in 1986 to produce a new alloy for orthopedic applications. It aims to use incompatible elements by optimizing properties such as flexibility and strength in implantation.

α+β titanium alloy has higher hardness, notch properties and strength than β-Ti alloy. β-Ti alloys containing biocompatible elements such as Zr, Nb, Mo, Ta or Fe have an elastic modulus closer to the bone than Ti6Al4V alloy. α+β alloys such as Ti5Al2.5Fe and Ti6Al7Nb have been used in studies to cover the negative effects of vanadium. Since these α+β alloys show mechanical and tribological properties close to Ti6Al4V alloy, it has been determined that the problem persists in applications due to the aluminum content. The high elastic modulus causes stress concentrations in the bones. The elastic modulus of the titanium alloy Ti6Al4V is lower than that of stainless steel and cobalt alloys. However, even in this way, the elastic modulus of the Ti6Al4V alloy is higher than that of bone. β-phase Ti alloys such as Ti15Mo and Ti13Nb13Zr were produced due to lower elastic modulus values than α+β alloys. Junior et al. reported that the Ti15Mo alloy gave good cell viability and cytotoxic test results. Stated in their studies [62]. The use of titanium and titanium alloys in biomedical applications is given in Table 3.4.

Table 3.4 Titanium alloys for biomedical applications [63].

Titanium Alloys	Specification
Pure Ti	(ASTM F67-89) Grades 1, 2, 3 and 4
Ti6Al4V	(ASTM F136-84, F620-87) α+β type
Ti6Al4V	(ASTM F1108-88) α+β type
Ti6Al7Nb	(ASTM F1295-92) α+β type (Switzerland)
Ti5Al2.5Fe	(ISO5832-10): α+β type (Germany)
Ti5Al3Mo4Zr:	α+β type (Japan)
Ti15Sn4Nb2Ta0.2Pd:	α+β type (Japan)
Ti15Zr4Nb2Ta0.2Pd:	α+β type (Japan)
Ti13Nb13Zr	(ASTM F1713-96): low modulus close to β type
Ti12Mo6Zr2Fe:	β-type (US) low modulus
Ti15Mo:	β-type (US) low modulus
Ti16Nb10Hf:	β-type (US) low modulus
Ti15Mo5Zr3Al:	β type (Japan) low modulus
Ti5Mo3Nb:	β type (US) low modulus
Ti35.3Nb5.1Ta7.1Zr:	β-type (US) low modulus
Ti29Nb13Ta4.6Zr:	β type (Japan) low modulus
Ti40Ta, Ti50Ta:	β type (US) high corrosion resistance

3.5 EQUILIBRIUM PHASE DIAGRAM OF TI-15MO ALLOY

The metastable Ti-15Mo alloy, type β, was developed around 1950 for the chemical industry [10]. Some decades later, the Ti-15Mo alloy, due to its mechanical properties associated with biocompatibility, was inserted as material for use in surgical implants with defined standards (ASTM F 2066) [64].

With a 3 to 20 wt% molybdenum content, the Ti-Mo alloy was analyzed for the phases present and crystalline structure, depending on the Mo content and mechanical properties (hardness, elasticity module, and flexural strength). The phases and microstructure results for CP-Ti and alloys from 3 to 20% of Mo in Table 3.5.

The pure Ti and alloys with Mo content in the range of 3.2 to 4.5 wt.% transformed from β phase-field (900°C) to martensite during water quenching. For pure Ti, the β phase transformed into an α' phase with a hexagonal structure. For Ti-3.2Mo, β phase transformed to α'' phase with an acicular martensitic microstructure, which has an orthorhombic form. For Ti-4.5Mo, β phase transformed to α'' phase and a small amount of ω phase. The martensitic structure changes from a hexagonal structure to an orthorhombic phase at a Mo content of around 3.1wt. %.

Table 3.5. Phases and crystalline structure of Ti and alloys Ti-Mo [65].

Mo %	Phase	Crystalline structure
CP-Ti	α'	Hexagonal
3-5	α'	Hexagonal
6	α'/ α"	Hexagonal/ orthorhombic
7.5	α"	orthorhombic
9	α"/β	Orthorhombic/cubic
10-20	β	cubic

As shown in Figure 3.6, when the Mo content increases to c2, the ω phase cannot be detected at room temperature. So, Ti-12Mo is dominated by the β phase. Also, the ω phase is formed in the metastable β type alloys with different Mo contents and disappears in Ti-Mo alloys with high Mo content [66].

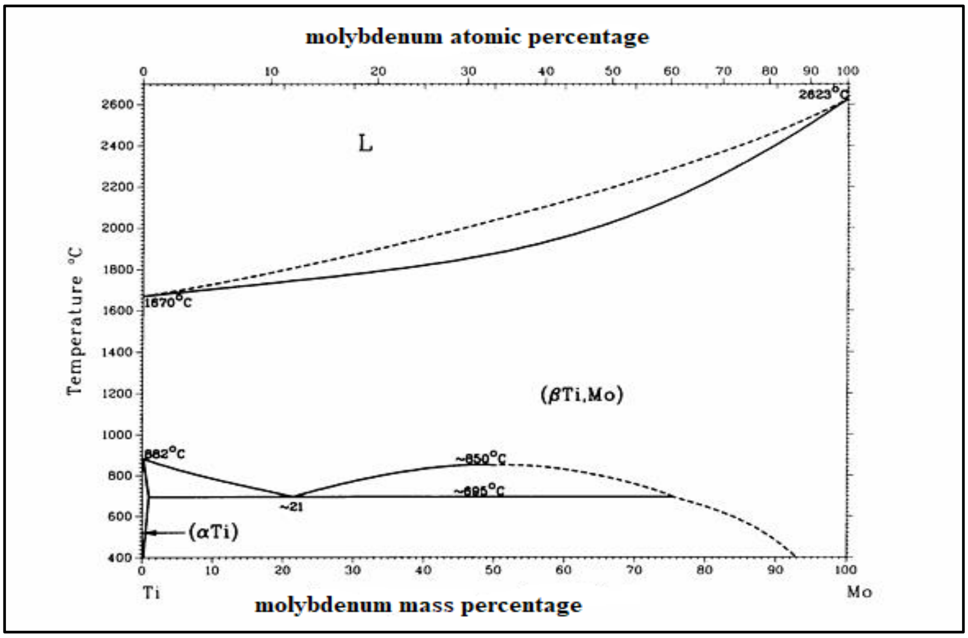


Figure 3.6. Phase diagram of Ti-Mo alloys [66].

The Ti-15Mo alloy, worked (annealing), is the most malleable alloy since its (reduction of Area) RA is the highest (82%) and, as expected from β alloys, it has a low elastic modulus (E) (78 GPa), and closer to the E of the human cortical bone (10 to 30 GPa); thus confirming the greater ductility and an excellent E value of the Ti-15Mo alloy, among the alloys fair use as a biomaterial [67].

The microstructure, residual porosity, mechanical properties, and cytocompatibility of Ti-Mo, developed via powder metallurgy, were investigated. The results show that homogeneous solid solutions were obtained with a low residual porosity rate for Ti alloys. Ti-20Mo, Ti-40Mo, and the Ti-60Mo alloy also presented a homogeneous solid solution but much higher porosity and was evaluated in about 26% to 28% of the samples. The compression tests showed that the alloys are very ductile compared to the Ti-6Al-4V alloy. Continuous layers of tissue were visible on each surface of the treated samples, indicating excellent cytocompatibility. The Ti-60Mo alloy had more outstanding cell adhesion due to its relatively high porosity. This cell adhesion is highly desirable in implants [67].

3.6 EFFECTS OF ALLOYING ELEMENTS FOR TITANIUM ALLOYS

Titanium and its alloys are frequently used materials in biomedical applications; It has full inertness in body tissue, adaptability to bone and other tissues, low density, high corrosion resistance and high strength, and has elastic modulus values close to the bone. Titanium alloys are used to diagnose and treat biomedical applications with different elemental additives. Most pure Ti and Ti6Al4V alloys, as well as molybdenum (Mo), palladium (Pd), tin (Sn), vanadium (V), aluminum (Al), niobium (Nb), tantalum (Ta) and zirconium (Zr) elements are added and used in applications [68].

In recent years, titanium and its alloys have been widely used in dentistry and medical applications. Typically, titanium was widely used in aerospace, marine and space applications. Due to its low density, durability and robust structure, and high heat and corrosion resistance, it increases its use in medicine and dentistry. The use of titanium in biomedical products has increased in recent years with the development of processing techniques. Today, titanium alloys are used in medical equipment such as dental implants, splints, crown bridges and partial dentures, stents and connectors, and joint prostheses. The alloying element additive improves the mechanical properties of titanium. Titanium alloys are classified into two groups according to which element is doped. Titanium alloys with element additions that will contribute to increasing or decreasing the transformation temperature are separated as α or β stabilizing elements. These alloying elements are classified as neutral, α or β stabilizers. Figure 3.7. shows the effects of some alloying elements on the phase diagram of titanium.

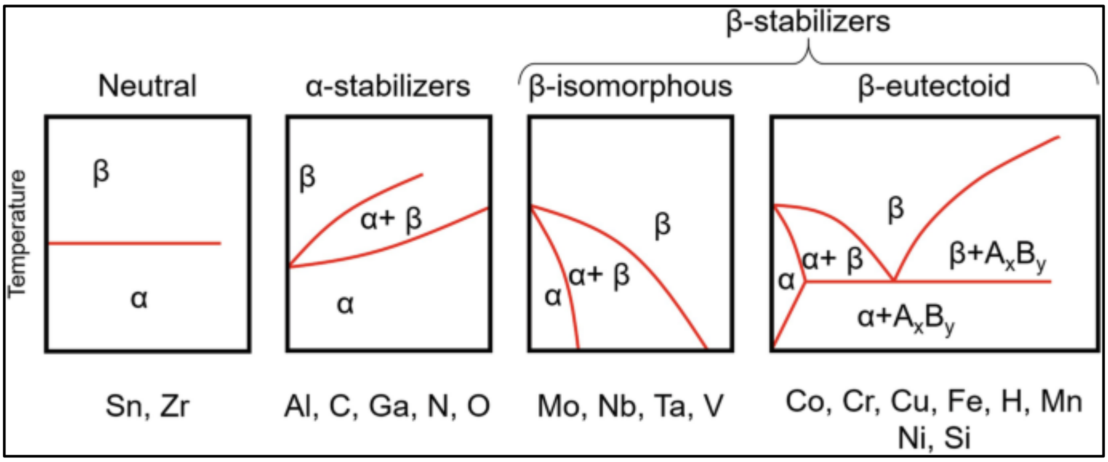


Figure 3.7. The effect of some alloying elements on the titanium phase diagram [68].

While the α-titanium region is increased by the α phase stabilizing elements, the β-titanium region and β phase stabilizing elements decrease to lower temperatures. The α-titanium phase stabilizer is Al, O, N and C. The most preferred element in doping is aluminum. When the research is examined, the alloying ratio of the Al content in the alloy must be at the limit of 6% not to form Ti3Al precipitate. Oxygen, on the other hand, increases alloy strength and fabricability for the α-titanium phase [51].

There are β-titanium phase stabilizing elements in two groups, β isomorphic and β eutectic. β-titanium isomorphic stabilizing elements; Nb, V, Mo, Ta and Re are the elements. The elements are β-titanium eutectic stabilizing elements; Cr, Si, Fe, Cu, Mn, Pd, Bi, W and Ni. On the other hand, Zr and Sn have neutral behavior [51]. Figure 3.8 shows the phase diagram alloyed with Mo element, where α and β titanium stabilizing structures are present. The mechanical properties of different titanium alloys are given in Table 3.6.

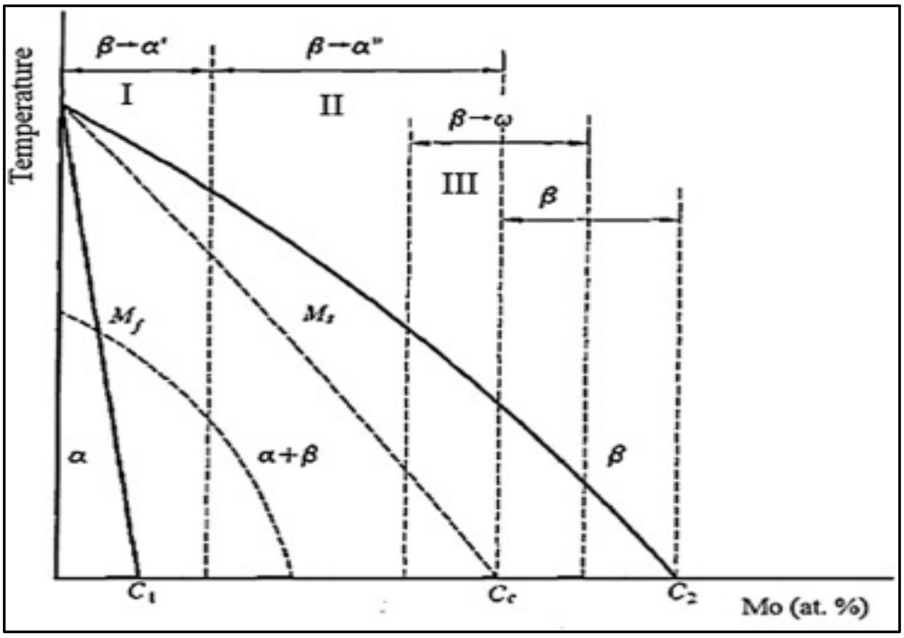


Figure 3.8. Alloying of titanium using Mo element in the triple phase diagram [66].

Table 3.6. Interstitial element limits and mechanical properties for CP Ti (Grades 1-4)[69].

Grade	O (max)	N (max)	H (max)	Yield stress (MPa)	Ultimate tensile strength (MPa)	Elongation %
1	0.18	0.03	0.015	170	240	24
2	0.25	0.03	0.015	275	345	20
3	0.35	0.05	0.015	380	450	18
4	0.40	0.05	0.015	483	550	15

3.6.1. Molybdenum

According to Bania [70] and Ho and Lin [65], a minimum of 10 wt% of the isomorphous β-stabilizing element is needed to stabilize the β phase for a Ti-Mo alloy at room temperature. Below this percentage, the alloy consists of a martensitic α″ phase that has a lower hardness than β-Ti-Mo. Ti–10Mo was tested to have the highest bending strength, and Ti–15Mo has the lowest modulus among the βTi-Mo and even lower than other alloys such as Ti–6Al–4V, Ti–6Al–7Nb, 316 L stainless steel, and Grade IV cp-Ti [71], due to a fine grain bcc structure was obtained. Such an alloy was claimed to have better processability. Indeed, such a metastable β phase Ti–15Mo alloy, manufactured by rapid quenching, has been marketed and sold for the orthopedic implant by Synthes USA. However, a careful selection of the concentration is necessary since the existence of ω phase at a low concentration of Mo (<15%) might have low-temperature ω → α transformation and thus affect the strength of the material [70].

3.6.2. Indium

For a long time, Indium (In) has been used in Pd- and Ag-based porcelain-fused-metal (PFM) applications. In fact, during the firing process of porcelain, the indium oxide film would be formed on the metal surface, which could be served as a 'bonding agent' between metal and porcelain [72, 73]. Also, cytotoxicity tests revealed that dental alloys that contain indium are safe. Therefore, using indium as an alloying element to improve the alloy to enhance cp-Ti's clinical performance was deemed reasonable.

Several studies on experimental Ti-In alloys have shown that Ti-In alloys were biocompatible. Further, adding indium to Ti could improve clinical performance regarding mechanical properties, corrosion resistance, and biocompatibility of dental implants [74]. According to a study regarding an alkali-heat treated Ti–In–Nb–Ta alloy, the surface analysis revealed good bioactivity.

For binary Ti-In alloys, Wang [75] found that the passivation current densities in artificial saliva solutions for Ti–In alloys and cp–Ti exhibited the same order of magnitude. Furthermore, Ti–10In and Ti–15In (10 and 15) denote the respective Indium wt% showed a transpressive behaviour and lowered current densities at high potentials under NaF.

Han et al. [76] have shown that Ti-In alloys (5–20 wt% In) not only exhibit a similar corrosion resistance to cp-Ti by electrochemistry but even a superior oxidation resistance compared to cp-Ti was revealed in Ti-In alloys. Therefore, Ti-In alloys might give good or better corrosion resistance than cp-Ti. Therefore, alloying of indium to Titanium effectively fabricated a new alloy that might have better mechanical properties without compromising its corrosion behaviour and cytocompatibility.

3.7. POWDER METALLURGY

Powder metallurgy (PM) is a technique for transforming metallic powders through compaction and consolidation of the part by controlled heating. It has well-defined steps: powder production, mixing / homogenization, compaction and sintering. MP offers a viable tool for producing complex components, producing parts with complex geometry and dimensional tolerances close to the final sketch. This technology uses approximately 90% of the raw material, with the possibility of avoiding or limiting the machining process, producing components with a good surface finish 94. Another essential feature is the ability to precisely adjust the chemical composition, densification and microstructural homogeneity [77]. Factors contribute to a reduction in production costs when compared to conventional processing. This particularly applies to relatively expensive materials such as titanium alloys. The high cost of extracting and manufacturing titanium makes it difficult to be widely used for conventional engineering, aerospace and medical implant applications. This generated the need for viable alternatives for producing titanium at low cost, with PM presented as the most economical [78]. Two techniques widely used to obtain titanium alloys are Pre-alloyed (PA) and Elemental Blended (BE). PA uses pre-alloyed powders that can be obtained, for example, by atomization, with subsequent compaction using hot isostatic pressing. However, it is a costly method despite producing components with excellent mechanical properties. The BE technique consists of mixing elementary powders, without the high costs mentioned in the first process. The subsequent steps are cold isostatic pressing, followed by sintering or uniaxial hot pressing. The elemental mixture is characterized by a low density of the products, about 95% of the theoretical density, while PA produces completely dense components.

However, with the advances in the BE technique, it is already possible to produce products with 99% of the theoretical density through hydrogenated powders [78]. Hydrogenation consists of a step of saturation of hydrogen atoms in the crystalline network of the material through controlled heating. Hydride particulates are obtained by comminution (generating a wide distribution of particulates) or the milling process (obtaining more significant control of particulate distribution). The removal of hydrogen from the material consists of the dehydrogenation step that occurs during heating under a vacuum [79]. The chemical reaction of this process can be written as M + H2 ← → MH2, where (M = metal), the two arrows symbolize that the reaction is reversible. The pressure of the hydrogen gas determines the direction of the reaction; if the pressure is above the equilibrium pressure, the reaction proceeds to the right, forming a metal hydride; if the pressure is below the equilibrium pressure, the metal hydride dissociates. The subsequent steps in the process of obtaining powders are compaction and sintering. The first consists of shaping the powders, and the second is the final consolidation of the part. Sintering is matter transport in a mass of powders or a porous, heat-activated compact. It results in a decrease in the specific free surface due to the growth of contacts between particles, the reduction of volume and the geometric alteration of the pores. There are two lines of sintered products of interest to the industry. One focuses on densifying and increasing strength without necessarily introducing dimensional changes. Another line is the production of elements with minimized densification, that is, with controlled porosity. Some parameters are necessary to satisfy the quality of the sintering process, such as the controlled conditions of heating and cooling speed, ambient atmosphere, temperature and temperature [80].

3.7.1. Powder Preparation (Mixing)

Powder preparation, an essential step after powder production, is the case of mixing the powders homogeneously. Metal powders are mixed with alloying elements and lubricants to obtain a homogeneous mixture. According to the weight of the part to be produced, the mixture is determined by the weight of the additives as a percentage. Between 0.5 and 1.5% of lubricant can be added to the structure. Stearic acid, metallic and zinc stearates, and paraffin are the lubricants most commonly used. Binders such as polyvinyl alcohol and paraffin are mostly used to assemble spherical powders. The main objectives of the lubricants are to allow the powders to move easily during pressing and to reduce the friction between the powder surface and the mold walls. Different mixer types are used in the mixing process in Figure 3.9 [81].

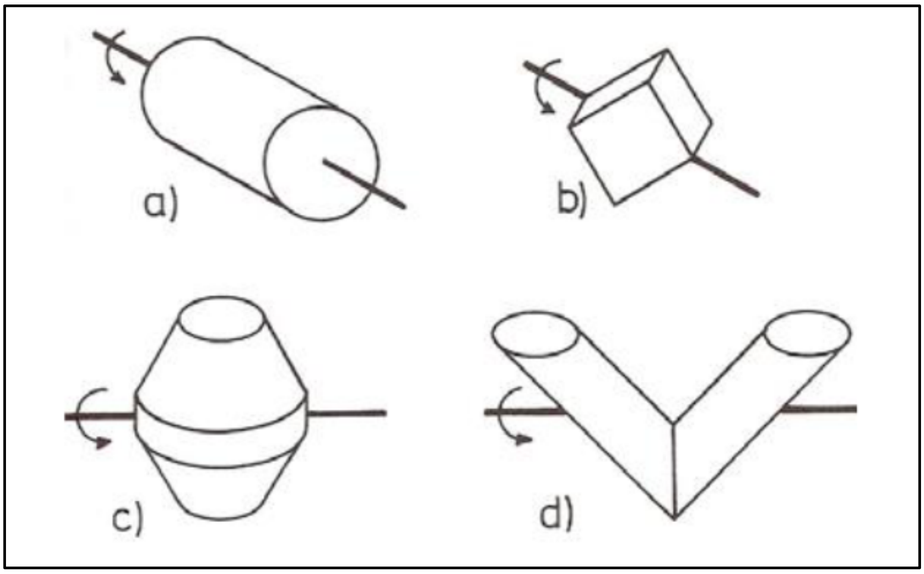


Figure 3.9 Mixer types used in mixing of powders [81].

During this process, the mixer must have an occupancy rate of at least 30%. Mixing is done at low speed (up to 100 cycles) and for half an hour. The raw density and post-sintering density significantly affect the optimum mixing process. Improper mixing prevents the formation of large pores after the heat treatment by observing the mechanical locking phenomenon and preventing the mechanical properties from reaching the desired levels. Mixing at high speed causes the homogeneity to be adversely affected and plastic deformation to occur.

3.7.2. Pressing (Compression)

Pressing is when powders prepared before sintering in powder metallurgy are compressed to gain a certain form. Different methods are used for pressing powders in powder metallurgy. There are methods such as uniaxial pressing, cold and hot isostatic pressing, powder forging, etc. The method, which is uniaxial pressing, also known as moulding, is used in metal alloys produced by powder metallurgy, which requires mass production. The pressing process starts with adding lubricant to reduce or eliminate the friction between the powder particles and the mold walls and is completed by taking the powder into the mold with the help of upper and lower punches and taking the solid shape of the sample after certain pressure. As the name suggests, compression occurs with one-way pressing. There are three different types of uniaxial pressing. It is a single and double system and floating mold. In all of these types, die, and punch are used together and have the potential to produce a large number of samples. In bidirectional pressing, compression is performed by simultaneously applying pressure from both the lower and upper punches. It is a floating die-type pressing, where the die placed in the middle of the punch performs the punching process with the action of two punches simultaneously. Figure 3.10 shows how uniaxial pressing takes place. In uniaxial pressing, the material is processed in four steps. Pressing, which starts with the rearrangement stage, continues with local deformation. Uniform deformation follows local deformation, and finally, the pressing process is completed with body compression. The first stage, rearrangement, enables the movement of dust particles, filling large gaps. In local deformation, on the other hand, contact between particles on the surface is ensured. In the next step, uniform deformation, the powder hardens, and the hole collapses. Finally, with the body compaction step, the powder particles took on a hard and solid structure.

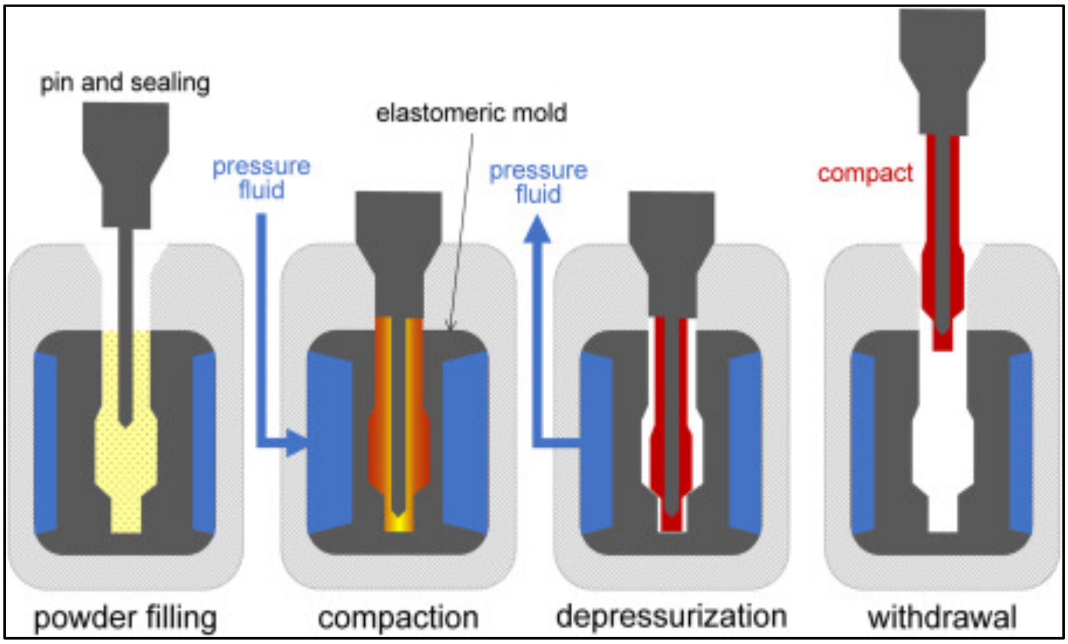


Figure 3.10. Uniaxial cold pressing [82].

There are many choices in the realization of pressing. Heating the powder and the mold, soft and hard moulds, changing the pressure dwell time and strain rate, and lubricating the mold and the powder can be shown among the preferences. Generally, the preferred situation in hard mold applications (especially for hydraulic and mechanical presses) is to press the powder particles uniaxially and at room temperature [83].

3.7.3. Sintering

In powder metallurgy, sintering, one of the important and technical stages that give the material its physical and mechanical properties, is applied to raw materials to produce less porous or nearly 100% density materials. Sintering can be done by thermomechanical processes such as hot and hot isostatic pressing, extrusion or forging. If the sintering process of a material with one component number is to be carried out, the process occurs at 40% and 66% of the melting temperature. In case of an increase in the number of components, the sintering process occurs at or close to the temperature of the element with the lowest melting temperature. Due to these conditions, to gain physical and mechanical properties during sintering, the process is carried out with a vacuum system in atmosphere-controlled environments so that it does not oxidize and react. Sintering, a heat treatment that enables powders to come together at high temperatures, is divided into two different titles: solid and liquid. While solid-state sintering is carried out below the melting temperature of the elements participating in the structure, the liquid phase formation begins and continues in liquid-state sintering, as the name suggests [84]. Material transport is called the reduction of the surface area of the porous powder particles exposed to the pressing process after mixing and the increase of their contact areas, as well as the activation by heat, which causes a decrease in the pore volume and a change in the pore shape. The strength of the wet-density materials obtained after the pressing process is low. It show resistance while trying to be removed from the mold is its expected strength during pressing. Therefore, it is used to gain mechanical properties such as strength and density of materials by sintering at a point below the melting temperature.

The situation starts with the contact event during sintering and continues with the formation of the solid-state bond, the formation of a new grain boundary with the necking of the contacting powder particles and then the formation of the two particles as a single piece and this is defined as the double-sphere sintering model [85]. For example, if we look at titanium alloys, It is preferred that the temperature considered for sintering multi-component alloys such as Ti6Al4V and Ti6Al7Nb is above 0.75. This ratio is known as the ratio of the sintering temperature of the alloys to the melting temperature [86]. It is recommended to sinter hard alloys 1400 – 1600 ºC, refractory metals 2000 – 2900 ºC, porous bronze and bronze-like alloys 600 – 800 ºC, ferrous alloys 1000 – 1300 ºC and Ti6Al4V 1100 – 1350 ºC in an atmosphere controlled and vacuum environment for two hours [87]. Even if the pressed powder particles come into contact or are incorporated into the structure, each particle behaves independently. The realization of sintering directly causes the contact points to enlarge and subsequently decrease the surface area, the globalization of the pores and the reduction of their volume, and the reduction of errors such as dislocations. A physical bond is formed between atoms and ions, the same as the bond in the crystal lattice.

3.7.4. Titanium Powder Metallurgy

Titanium is a costly material that is not easy to manufacture for applications. Despite its negative aspects, titanium and its alloys have recently been preferred due to their superior properties. The inability to produce titanium and its alloys by casting causes problems such as post-production processing problems and unwanted structures appearing in the material. The solution is a production technique recommended for titanium powder metallurgy and its alloys to reduce the cost [87, 88]. If it is desired to produce titanium with powder metallurgy, the first great advantage is that additional processes such as heat treatment, machining and joining are not used for homogeneity in casting. In addition, limiting situations such as oxidation, impurity, and formation of unwanted structures (impurity) may occur during production with the powder metallurgy technique. Oxidation, which reduces the fatigue strength of the materials, is undesirable in the parts produced with this technique. Since titanium tends to react with oxygen, the most suitable conditions should be determined and mixing, pressing and sintering processes should be carried out respectively.

Considering the production systems, titanium powders are produced in the 40 to 150 μm particle size range. Generally, titanium alloys used in powder metallurgy are produced from pre-alloyed or elemental powders. Elementally produced titanium powders are of low value in terms of cost and properties compared to pre-alloyed powders. For example, if it is desired to produce a titanium alloy with high strength values, a piece can be produced with an angular structure and pre-alloyed powder particles in a hard mold with a single or double-acting press under 500 to 650 MPa pressure, with a density close to 90% before sintering. In order to achieve a density value close to 90%, it is expected to reduce the porous material by processing under 350 to 400 MPa pressure and to achieve the same mechanical property values in all directions by isostatic pressing [89]. Titanium and its alloys are produced with a vacuum system in atmosphere-controlled tube furnaces. It is recommended to provide the necessary equipment to evacuate the gases that arise during the sintering with the help of a vacuum, to increase the temperature, to reach high vacuum values and to wait up to 4 hours at 1250 ºC. In using inert argon gas as a protective atmosphere, it is recommended to wait 2 hours in the temperature range of 1000 - 1200 ºC [87]. Using the powder metallurgy technique, it is possible to manufacture and design materials with a porous structure and produce dense implants with high strength. Considering the bone structures, there are two different structures, porous and dense. The powder metallurgy technique ensures that the implants used in dentistry applications are produced in a porous structure with features suitable for bone. For example, Ti30Ta alloy is preferred in the cortical bone because it has an elastic modulus close to the bone and a 15-20% pore amount. Bone-compatible titanium alloys such as Ti15Mo5Zr3Al, TMZF, Ti13Nb13Zr, Ti6Al7Nb and TNZT are also produced by powder metallurgy technique and offered for use. The production of these alloys, which are compatible with bone, using powder metallurgy technique, leads to the formation of alloys with a wide range of product portfolios and mechanical properties by following different conditions for different situations. This situation enabled the production of parts with desired properties by preferring superior aspects of mechanical or physical properties in different applications.

3.8 BIOMEDICAL APPLICATION OF TI-15MO ALLOY

Ti-15Mo alloy is widely used in biomedical applications due to its excellent biocompatibility, corrosion resistance, and mechanical properties. Here are some of the biomedical applications of Ti-15Mo alloy:

1. Orthopedic Implants: Ti-15Mo alloy is used in orthopedic implants such as hip and knee replacements, bone plates, and screws. The alloy's high strength and biocompatibility make it ideal for load-bearing implants [90].
2. Dental Implants: Ti-15Mo alloy is also used in dental implants due to its biocompatibility and corrosion resistance. The alloy's mechanical properties make it suitable for dental implants requiring high strength and durability [39].
3. Cardiovascular Implants: Ti-15Mo alloy is used in cardiovascular implants, such as stents and heart valves. The alloy's biocompatibility and corrosion resistance make it suitable for use in these implants, which are in contact with blood and other bodily fluids [91].
4. Surgical Instruments: Ti-15Mo alloy is used in surgical instruments due to its high strength and corrosion resistance. The alloy's biocompatibility is also suitable for use in instruments that come in contact with body tissues [92].
5. Drug Delivery Systems: Ti-15Mo alloy is used in drug delivery systems due to its biocompatibility and corrosion resistance. The alloy can make implantable devices that release drugs over a long period [60].

Overall, Ti-15Mo alloy is a versatile material with several biomedical applications due to its excellent biocompatibility, corrosion resistance, and mechanical properties.

PART 4

MATERIALS AND METHOD

4.1 MATERIALS

In this study, Ti, Mo, and In powders were used to prepare the specimens for the sintering process, for the various tests, procedures, all of the high purity, irregularly shaped, manufactured by Lemandou Ltd. Co. China, as described in table 4.1.

Table 4.1. Raw materials properties

Powder	**Purity**	**Average partials size (µm)**	**Density g/cm^3**
Titanium	99.85	66.29	4.51
Molybdenum	99.98	18.67	10.28
Indium	99.99	27.29	7.31

Samples coded in Table 4.2 were prepared in the present work [93][93][93][93]. Figure 4.1 summarizes the overall program used in the present work.

Table 4.2 Prepared samples in the present study.

Sample No.	**Ti (wt.%)**	**Mo (wt.%)**	**In (wt.%)**	**Salt (vol.%)**
1	83	15	2	0
2	83	15	2	30
3	83	15	2	40
4	83	15	2	50
5	83	15	2	60

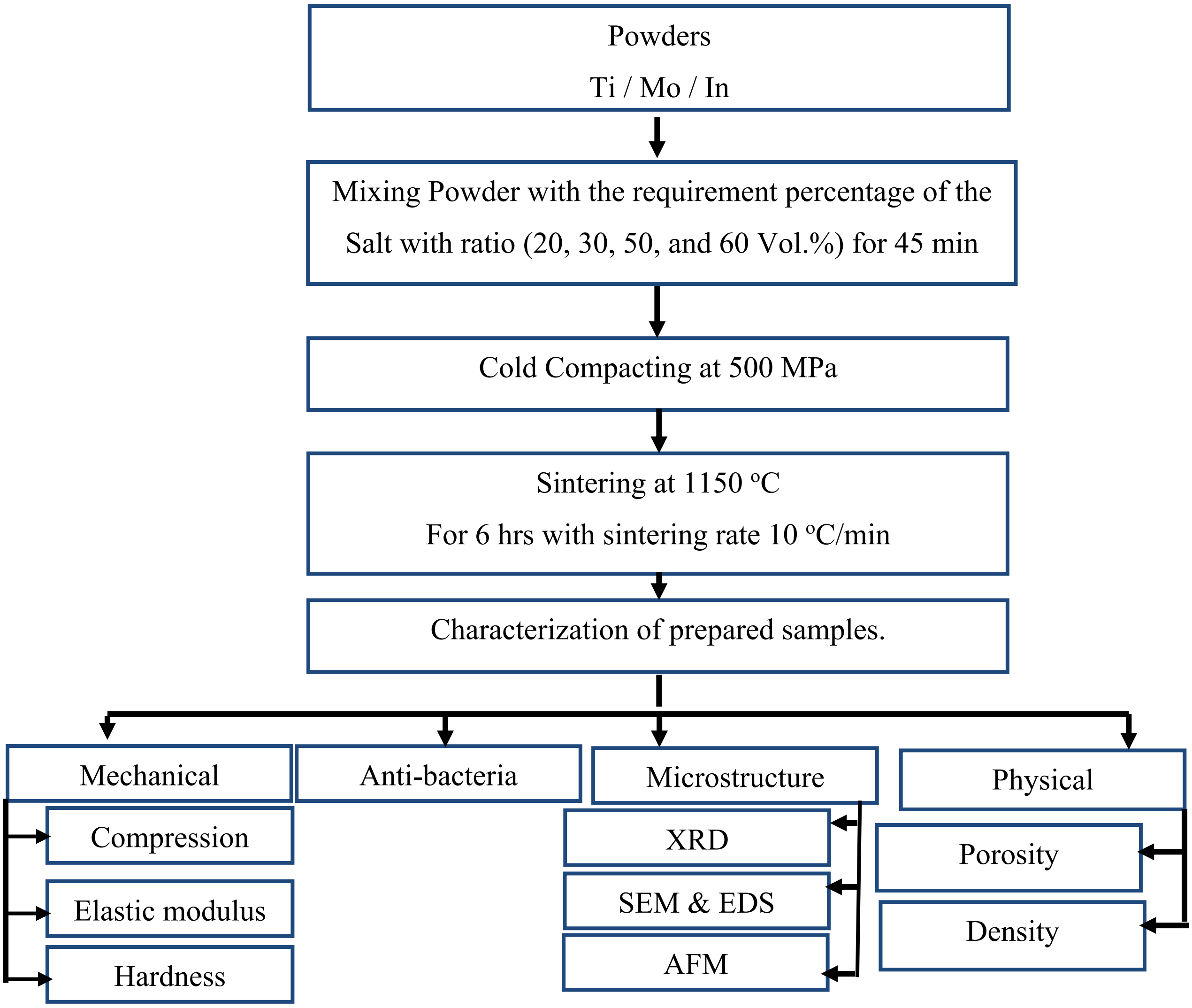


Figure 4.1. The experimental program of the present study.

4.2 PRODUCTION OF POROUS ALLOY

4.2.1. Mixing of Powder

Ti, Mo, In, and salt ammonium carbonate (NH_4HCO_3) powders were mixed using the Mixing Machine given in Figure 4.2 for a homogenous distribution of powder particles in the presence of 2 wt.% acetone (to prevent friction between the mold and powders particles and to reduce the oxidation of powder particles during mixing process). The samples have the following composition: Ti -15%Mo - 2%In, as shown in Table 4.2.

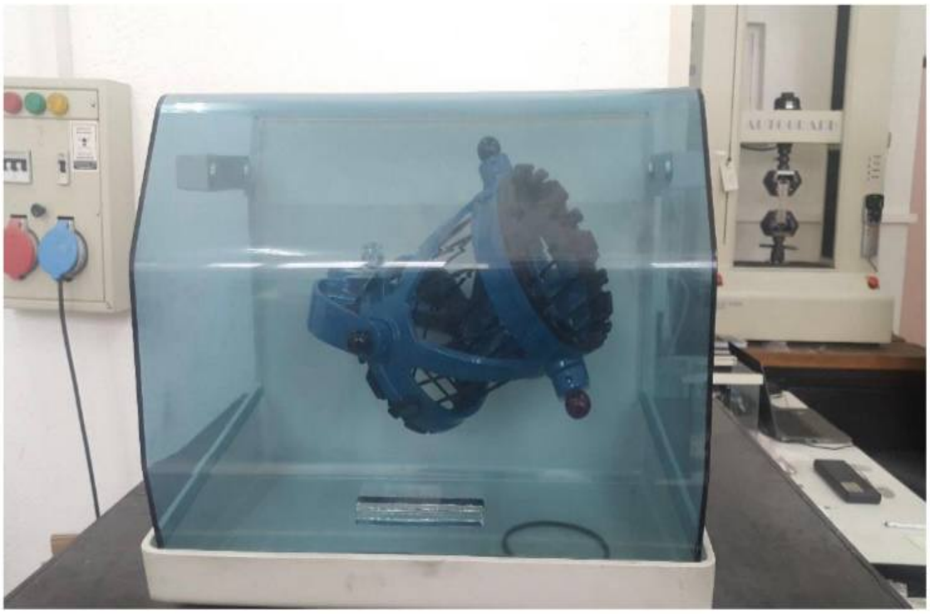


Figure 4.2. Turbula triaxial mixer.

4.2.2. Cold Pressing Process

The mixed powder was cold pressed into a mold with a diameter of Ø13x100 mm for the experimental studies, as shown in Figure 4.3.prrssing was performed by applying a load of 500 MPa in the hydraulic press shown in Figure 4.4 before sintering.

The samples had different sizes, according to the test that was carried out: for the compression tests, they had the dimensions of 13 mm in diameter by 26 mm in height (A1), and for the other tests, they had the dimensions of 13 mm in diameter by 5 mm in height (A2) (Figure 4.5).

Figure 4.3.Metal mold.

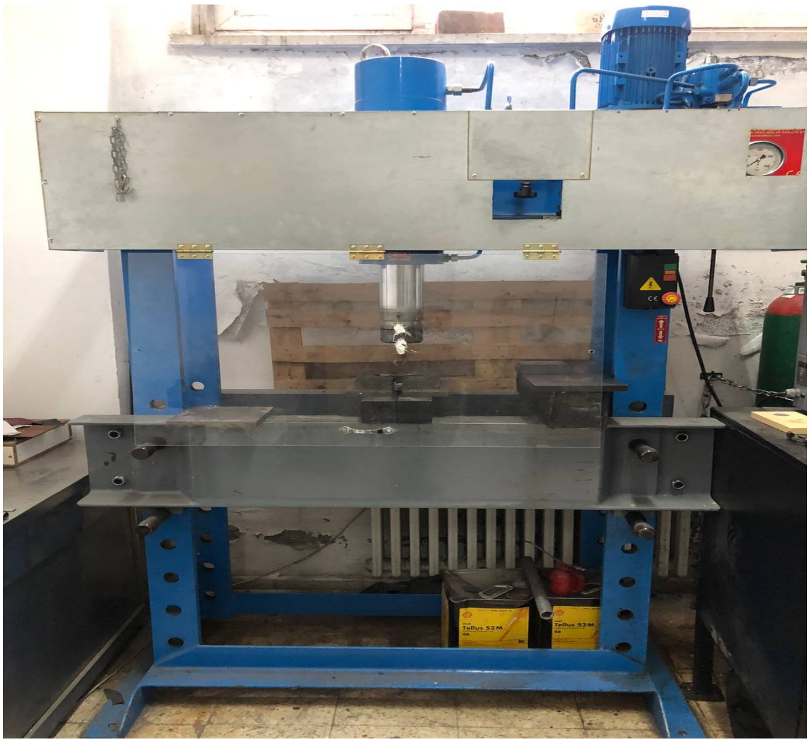


Figure 4.4. Hydraulic pressing.

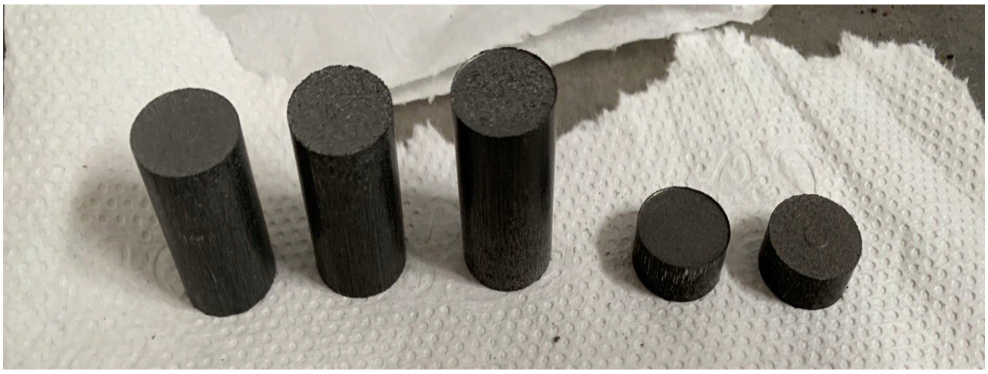


Figure 4.5. Different sizes of specimens (13 mm x 26 mm and 13 mm x 5 mm).

4.2.3. Sintering Process

After compaction, the specimens of titanium alloys are subjected to a sintering process to gain their mechanical and physical properties. The sintering process was carried out in the vacuum atmosphere-controlled tube furnace, shown in Figure 4.6, in an argon environment at 10 °C/min heating and cooling rates at 1150 °C for 6 hours. The samples are sintered on alumina substrates to avoid any reaction.

Figure 4.6 Atmosphere-controlled Tube Furnace used for the sintering process.

4.2.4 Metallographic Sample Preparation

Cold embedding was used to examine samples for metallography Figure 4.7. The Struers cutting equipment was used for cutting, and the sanding and polishing process was used for metallography. After this, all samples were produced using normal metallographic techniques and dispersed for 60-90 seconds with 95 ml ethanol and 5 ml nitric acid (Nital solution).

Figure 4.7. Cutting and grinding machine.

4.3 MICROSTRUCTURE CHARACTERIZATION

4.3.1. X-ray Diffraction (XRD)

The analyzes were carried out in a Rigaku Ultima IV X-Ray Diffraction Spectrometer, shown in Figure 4.8, which uses CuKα radiation. The parameters used were: voltage of 35 kV, current of 40 mA, 2θ angle ranging from 10 to 90° with a step of 0.02° and time of 1 second. The XRD analysis aimed to observe the sample phases sintered at three different temperatures. From the obtained results and formed diffractograms, the peaks were analyzed using sheets of the JCPDS (Joint Committee for Powder Diffraction Standards). After identifying the phases present and analyzing the intensity of the peaks, the relative and quantitative intensity of the phases present were calculated from the total intensity of the peaks.

Figure 4.8. Rigaku Ultima IV X-Ray Diffraction Spectrometers.

4.3.2. Scanning Electron Microscope (SEM)

After the sintering process, the microstructure and energy dispersive spectrum (EDS) was used for chemical and elemental analysis of the materials. The prepared porous titanium alloys were examined using the Carl Zeiss ultra plus Gemini FESEM model scanning electron microscope in the Karabuk University Scientific Technology Application and Research Center (MARGEM) laboratory (Figure 4.9).

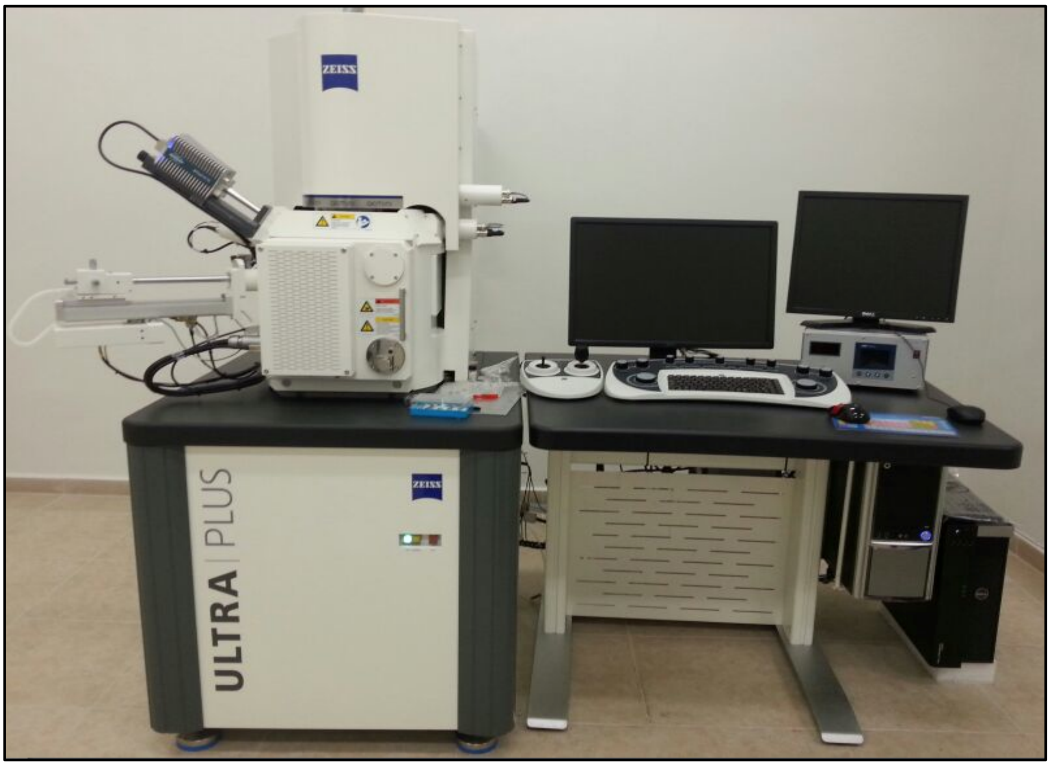


Figure 4.9 Carl Zeiss ultra plus Gemini FESEM.

4.3.3. Atomic Force Microscopy

To examine the surface's topography and determine its roughness, atomic force microscopy was carried out using a scanning microscope (NaioAFM 2022 Nanosurf Switzerland), as shown in Figure 4.10. A needle measuring in the micron range is moved over the surface scanned by the instrument. This needle is perpendicular to the holder and held in a horizontal holder. The laser beam is directed onto the holder, which rises and falls with the height and fall of the needle, and therefore with varying surface topography from high to low, and the receiver records the reflection of the laser beam on the holder. Consequently, the scanned surface's topography may be determined and mapped based on how the laser beam's reflection moves.

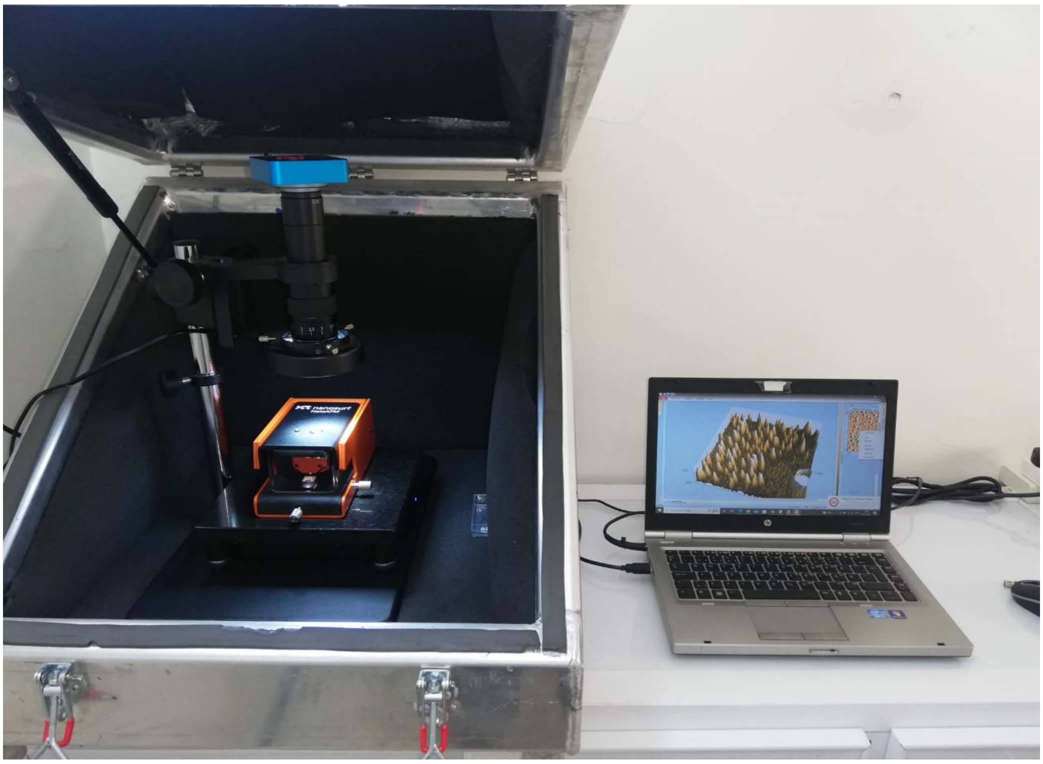


Figure 4.10. Atomic force microscopy.

4.4. PHYSICAL PROPERTIES

The Archimedes principle performed density measurements in the sintered samples following ASTM B327, for which a Mitutoyo digital calliper with 0.001mm sensitivity and a Scaltec scale with 0.0001g precision were used (Figure 4.11). The density of three samples from each group was calculated, and the average of these values was calculated according to equation 4.1.

The porosity of the sintered alloys, which influences implant materials' mechanical and biological properties at the bone-implant interface, means the fraction of the pore volume concerning the total volume and is determined by equation 4.2.

$$p = \left[\frac{B-A}{(B-F)*D^{O}} * 100\right] * D_{W} \text{........}(4.1)$$

$$\rho = \left[\frac{A}{B-D}\right] * D_{W} \quad \text{..................} \ (4.2)$$

Where:

D_W – The density of water = 0.9956 g/cm^3.

D^O - The density of oil= 0.634 g/cm^3**.**

Figure 4.11. Scaltec precision scale with 0.0001g precision.

4.5. MECHANICAL PROPERTIES

4.5.1. Ultrasonic Wave Test

Ultrasonic wave device type (cut-4) was used (Figure 4.12), and a digital monitor appears directly at the time of transfer of ultrasonic wave and can calculate the longitudinal and transverse velocities was used to investigate the mechanical properties (Elastic modulus and Poisson ratio) by applying the longitudinal and transverse velocities value on the equations below.

$$\upsilon = \frac{1-2(\frac{V_T}{V_L})^2}{2-2(\frac{V_T}{V_L})^2} \quad \ldots\ldots\ldots\ldots\ldots\ldots(4.3)$$

$$E = \frac{{V_L}^2\, \rho\, (1+\upsilon)(1-2\upsilon)}{(1-\upsilon)} \quad \ldots\ldots\ldots(4.4)$$

Where:

ν: Poisson's ratio, V_T: Shear (Transverse) velocity (m/sec), V_L: Longitudinal velocity (m/sec.), E: Elastic modulus (GPa), ρ: Density (kg/m^3).

It is about evaluating the mechanical behavior of the samples, in terms of stiffness, by the value of E, comparing the E of the implant with the E of the bone. E values of the implant lower or, at most, equal to the E of the bone allow load absorption by the implant at the bone-implant interface and consequent lower stress values. Especially for biomedical applications, the implant material's response must be considered carefully to avoid the so-called "stress shielding effect." This essentially corresponds to the reabsorption of the load by the bone, at the bone-bone implant interface and the consequent loosening of the implant, due to the difference (incompatibility) between the elastic modulus of the implant material and that of the bone. This is generally the case for implant materials much harder than bone.

Figure 4.12. Ultrasonic wave device type (cct-4).

4.5.2. Compression Strength Test

The compression tests were performed in a universal testing machine, brand Zwick/Roell z600, at a speed of 1mm/min (loading rate), as shown in Figure 4.13. The results of the limit stress were recorded yield strength (relative to deformation of 0.2%) and the ultimate compressive strength of the samples, which were obtained from stress-strain curves. The results of the compression tests were essential to determine the values of the yield strength and compressive strength and evaluate the mechanical response of porous samples under monotonic compressive load.

Figure 4.13. Monotonic compression testing machine.

4.5.3. Hardness Test

The hardness test was performed on the Qness hardness measurement in the MARGEM laboratory of KARABUK University (Figure 4.14). Appropriate grinding and polishing were done before subjecting the specimens to the test. The macrohardness Brinell test includes using load (187.5 kg/mm²) on the specimen to measure its hardness by a carbide ball diameter of 2.5 mm for 10 sec. The hardness was recorded as an average of three readings for each specimen.

Figure 4.14. Brinell Hardness device

4.6. ANTIBACTERIAL TEST

The agar well diffusion technique examined the materials' antibacterial effectiveness. Bacteria such as "Escherichia coli" and "Staphylococcus aureus" (both of which are gram-positive) were employed. The Chemicals used during the process.

• Müller Hinton Agar
• Tryptone Soy Broth
• Blank Antibiotic disc
• Cefoxitin 30 µg
• Penicillin G 10 µg

The test substance was applied directly on the agar plate by opening the same amount of space with the notification of the person requesting the test; no pre-treatment was applied. After the liquid culture (1.5 x 108 CFU/mL) is homogenized by shaking, it is immersed in the swab, wetted, and sowing is done as indicated below.

• It is spread over the entire surface, starting from one side of the agar.
• The exact process is repeated by turning it to 60 degrees.
• The exact process is done with the same swab by turning it 60 degrees again.
• For the fourth time, the swab is applied to the surface of the agar on the petri dish, and the process is completed. Wait 5-10 minutes at room temperature for the surface of the agar to dry. Discs with antibiotic as positive control and discs with 10 µl test material added are placed on the medium and incubated at 37°C for one night. Zone diameters formed after one night of incubation were measured with a calliper .

PART 5

RESULTS AND DISCUSSION

In this part, the characterization of prepared alloys Ti-18Nb-2In with and without different salt concentrations by XRD, SEM/EDS analysis and the exam of surface morphology will be scanned by AFM analysis. Mechanical properties such as tensile, compression, hardness and antibacterial behavior of experimental specimens were also measured to show the suitability of these alloys for biomedical applications.

5.1. MICROSTRUCTURE CHARACTERIZATION

5.1.1. X-ray Diffraction (XRD)

As intermetallic compounds and a phase change may also affect the mechanical characteristics of the referenced alloy [57]. X-ray diffraction analysis was the primary method used to examine the microstructure of the Ti-15Mo-2In alloys. Titanium does not undergo a phase shift under equilibrium circumstances; only this is classed as Ti [94]. If more reactive beta elements are not present, the phase may become metastable and thermodynamically more stable than the martensitic phase upon rapid cooling [57].

The XRD results of bulk Ti-15Mo-2In alloy after sintering at 1150°C are given in Figure 5.1. α-Ti, β-Ti, and Ti_3In_4 peaks were observed in XRD results. The β-Ti peaks were observed due to Mo being the β stabilizing element besides. Ti-15Mo-2In alloy, when sintered at 1150°C, forms α- Ti phases, as is well known. Ho et al. [65] found that the alloys containing 9% Mo had the coaxial phase, while 10% Mo contained the dominant phase. Following the previous finding [76], additional research has shown that alloys containing 15% Mo exhibit a dominating β phase; it is well-known that Mo is a crucial phase stabilizer [95, 96].

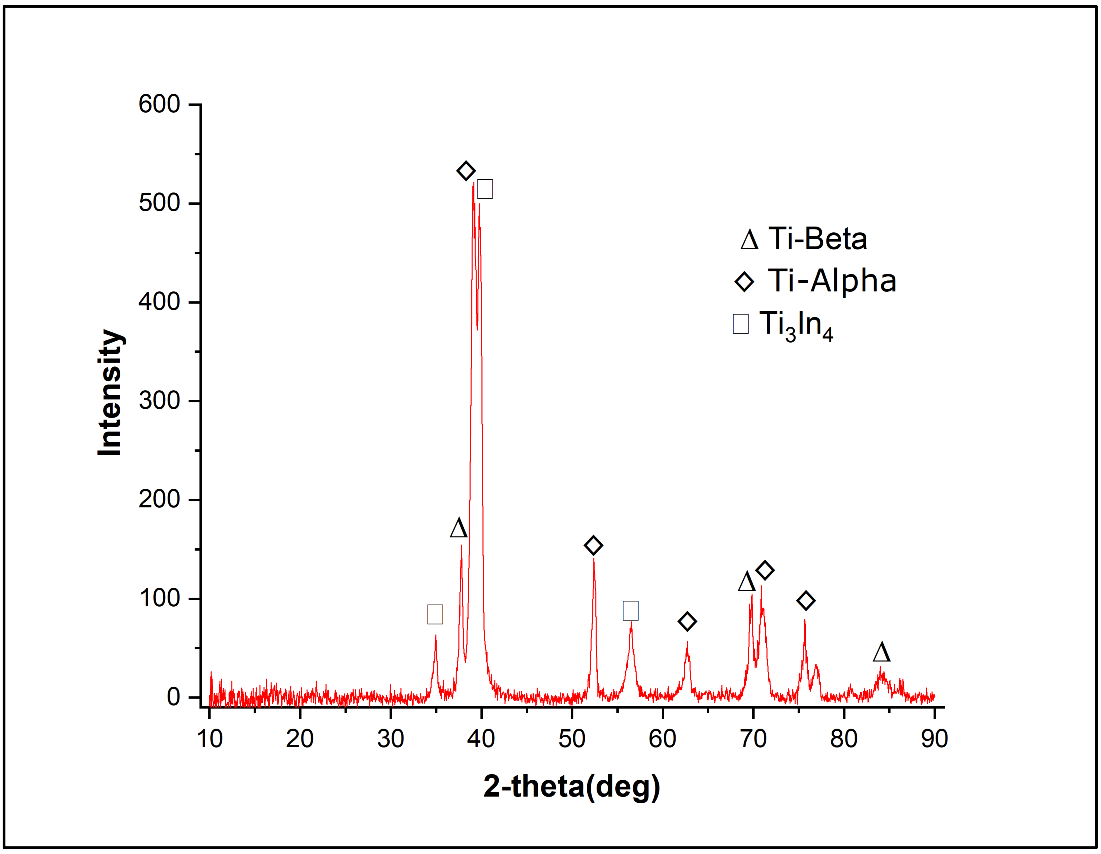


Figure 5.1. XRD result of Ti-15Mo-2In alloy after sintering at 1150°C.

5.2.2. Scanning Electron Microscope (SEM)

The microstructure of the etched samples was obtained using a scanning electron microscope (SEM). The microstructure and grain boundaries have been scratched out of the specimens. Different-sized pores are irregularly shape . Crystalline grain and solid-phase boundaries in multiphase alloys are metallic materials' most general microstructure characteristics [97].

As the microstructure of sintered specimens shows a multiphase structure, with the β-Ti phase correlating to the XRD result [98], SEM pictures are sensitive to chemical composition. In Figure 5.2, grain boundaries and varying sizes of pores were visible in etched alloys' SEM images.

Roughness improves stability by negatively impacting osteoblastic cell growth during cell identification and adhesion processes [98]. The relief areas have the roughest surfaces.

Since the salt evaporates during sintering, the XRD patterns and SEM pictures in Figure 5.3 reveal that adding salt did not affect the produced phases and structure of the created alloys.

The space holder technique is attractive and Sodium chloride (Ammonium carbonate (NH_4HCO_3)) is often used as the space-holding substance because it is inexpensive, highly soluble in water, generally non-toxic, commercially available, and easily removed utilizing solvent-debonding procedures. Ammonium carbonate (NH_4HCO_3) residues also do not affect performance in vivo [99]. Nevertheless, the processing must remove the crystal water before it can be utilized; otherwise, it would explode and disintegrate at a specific temperature. When it breaks down, ammonium carbonate (NH_4HCO_3) fills the voids of the ready-made mold, causing it to warp. Desiccators preserve the dry ammonium carbonate (NH_4HCO_3) particles until required.

Certain researchers have effectively manufactured porous titanium using plasma treatment [100], but removing the post-process is essential to lowering the entire process cost. Powder injection molding and space holders are two methods that may be used to create porous titanium.

In order to have a more evident study of the microstructure and, thus, to get crucial information, Scan electron microscope SEM images were always chosen as the principal kind. Elemental dispersion spectra (EDS) of Ti-15Mo-2In with and without salt alloys after sintering at 1150°C are given in Figures 5.3 and 5.4. When the spectrum analysis results were examined, the spectrum analyses taken from different regions showed that the alloy was homogeneously distributed and that the alloys obtained spectrum results close to the additive amounts. It also coincides with the EDS results, where the microstructure is homogeneously distributed. Figures 5.3 and 5.4 show that the β-Ti peak is more prominent than the other elements in the EDS phases for the manufactured Ti-alloys. Furthermore, Molybdenum (Mo), Indium (In) were also found.

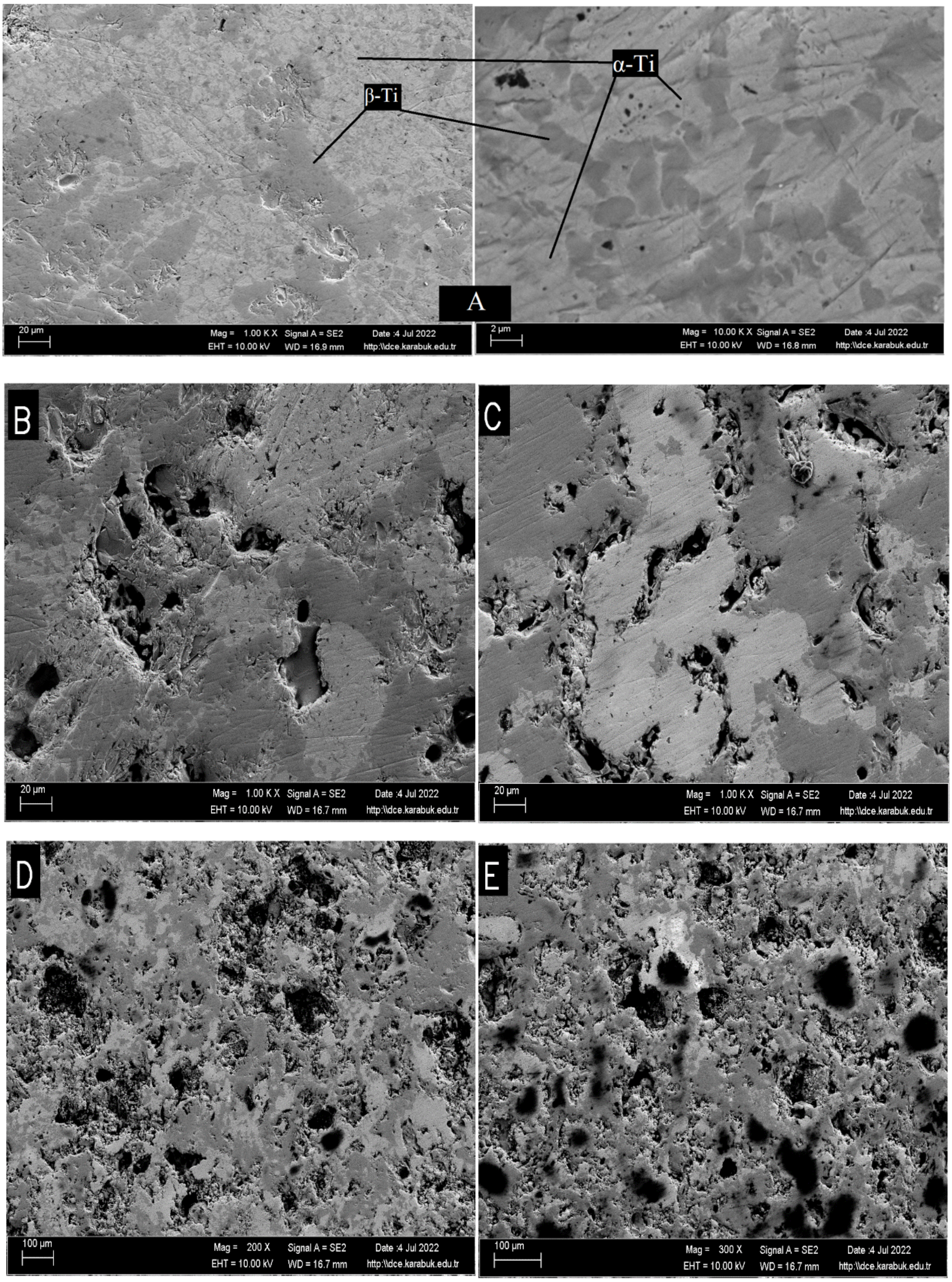
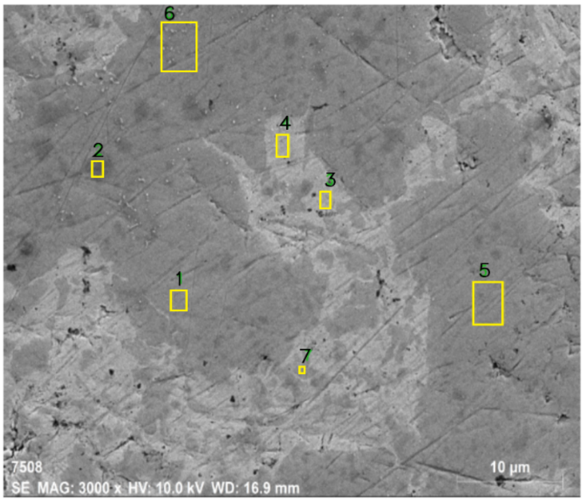
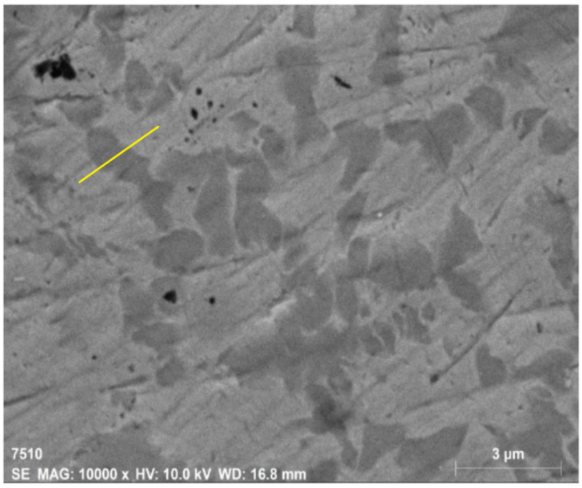


Figure 5.2. SEM of A)Ti-15Mo-2In, B) alloy with 20%salt,c) alloy with 30% salt, D) alloy with 50% salt, and E) alloy with 60% salt.



Mass percent (%)

Spectrum	Ti	Mo	In
1	95.82	0.64	3.55
2	96.88	1.25	1.88
3	70.45	24.89	4.66
4	70.35	24.81	4.84
5	99.30	0.70	0.00
6	98.98	0.95	0.07
7	92.19	4.01	3.80
Mean value:	89.14	8.18	2.68



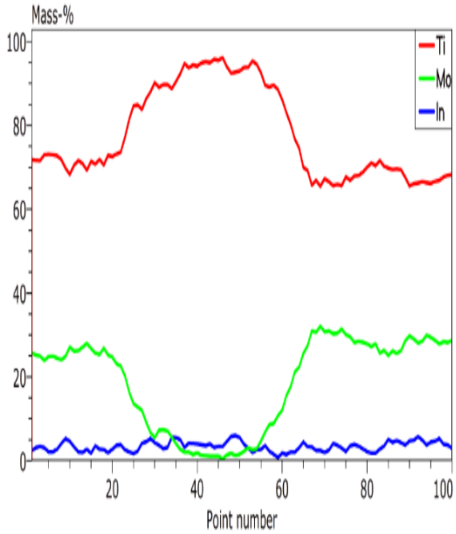
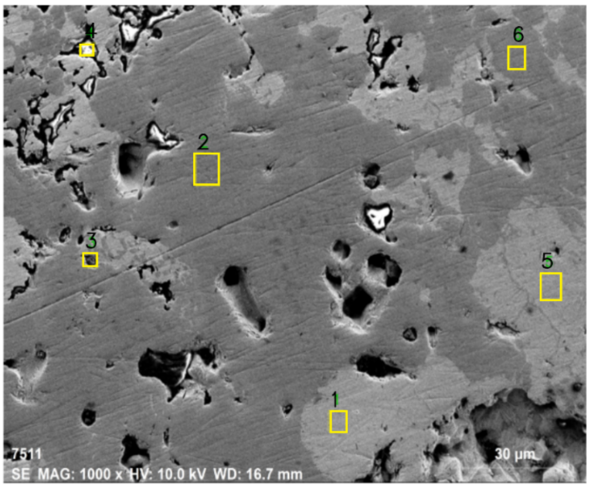


Figure 5.3. EDS for β-Ti-15Mo-2In



Mass percent (%)

Spectrum	Ti	Mo	In
1	63.11	35.26	1.62
2	99.24	0.46	0.30
3	75.72	19.51	4.76
4	45.18	42.55	12.27
5	71.20	26.17	2.63
6	95.63	0.56	3.81
Mean value:	75.01	20.75	4.23

Figure 5.4. EDS for β-Ti-15Mo-2In with 20 wt.% of salt.

5.2.3. Atomic Force Microscopy

Atomic force microscopy (AFM) analysis of single profile lines (2D) may provide insight into a surface's topography. When there are many peaks and valleys in the images, the average peak-to-valley difference (PVD) value is more accurate than the arithmetic means height (Ra) and root-mean-square height (Rq).

Figure 5.5 shows the morphology of the base Ti-15Mo-2In alloy, which has dimensions in the micrometre scale and a flat scan in the extracted profile due to the larger scale of the particles, and Figures 5.6 through 5.9 show the surface morphology of Ti-15Mo-2In alloys with different concentration of salt, which exhibits a distinct peak-to-valley difference at various length scales. Measurements of the surface's horizontal or lateral properties establish the dimensions of the distance between items. Generally, a spacing parameter will have features like peaks and valleys.

Vlcak et al.[101] investigation of the morphology of Ti-Nb-Zr-Ta alloy revealed the presence of pores, scratches, and tiny dimples, which may represent defects stemming from the material's random heterogeneity.

The results are consistent with the typical roughness attribute measured by AFM. Several studies have demonstrated that better bone-to-implant contact is achieved when the implant's surface roughness is between 1 and 10 μm. Bone does not make as good a contact with titanium that has been polished [102]. It must be abrasive enough so Ti can integrate with the bone.

Table 5.1 lists the surface properties of the prepared porous alloys. The Sa value corresponds with the Ra value measured with a contact profilometer. It's worth noting that the topographical profile is virtually perfectly Gaussian. These characteristics define a flat, even surface devoid of peaks and troughs of any significant size. As will be shown shortly, this is important since it promotes fibroblast attachment. Figure 5.5-5.9 shows the Abbott-Firestone curve for prepared porous alloys. The curved line shows the probability Amplitude Density Function; it shows how likely it is to locate a point on the surface at each of the given heights and where the mean line lies. As was predicted due to the existence of the grooves, it verifies a nearly balanced ratio between positive and negative characteristics on the surface, with a slight preponderance of valleys.

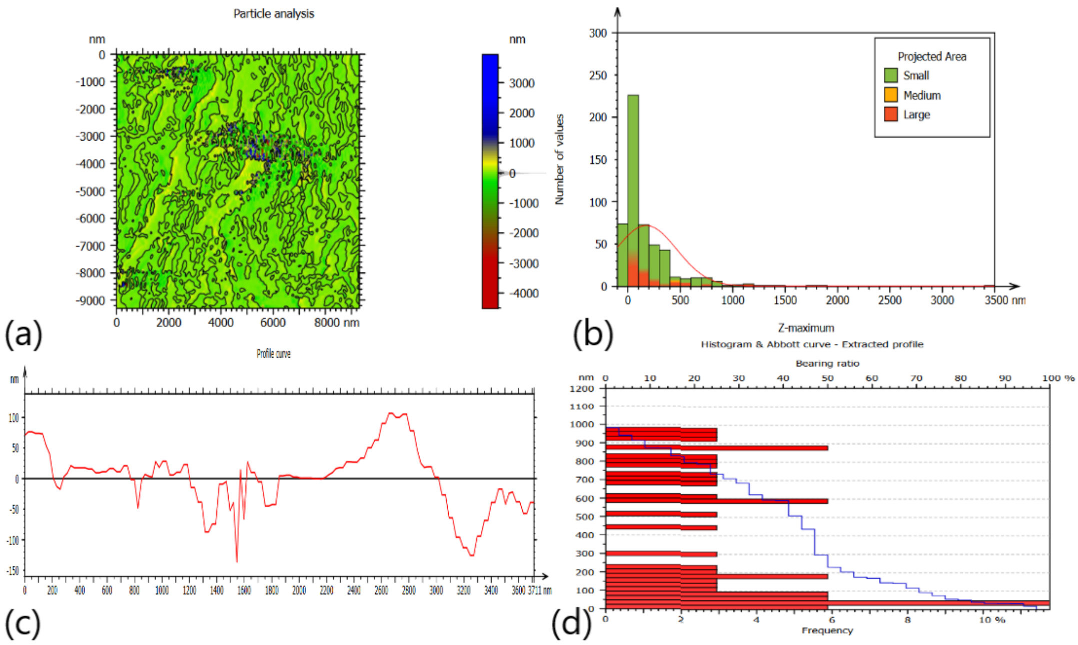


Figure 5.5. AFM analysis of Ti-15Mo-2In alloy a) Particle analysis, b) Particles Threshold, c) Profile curve, and d) Histogram & Abbott curve.

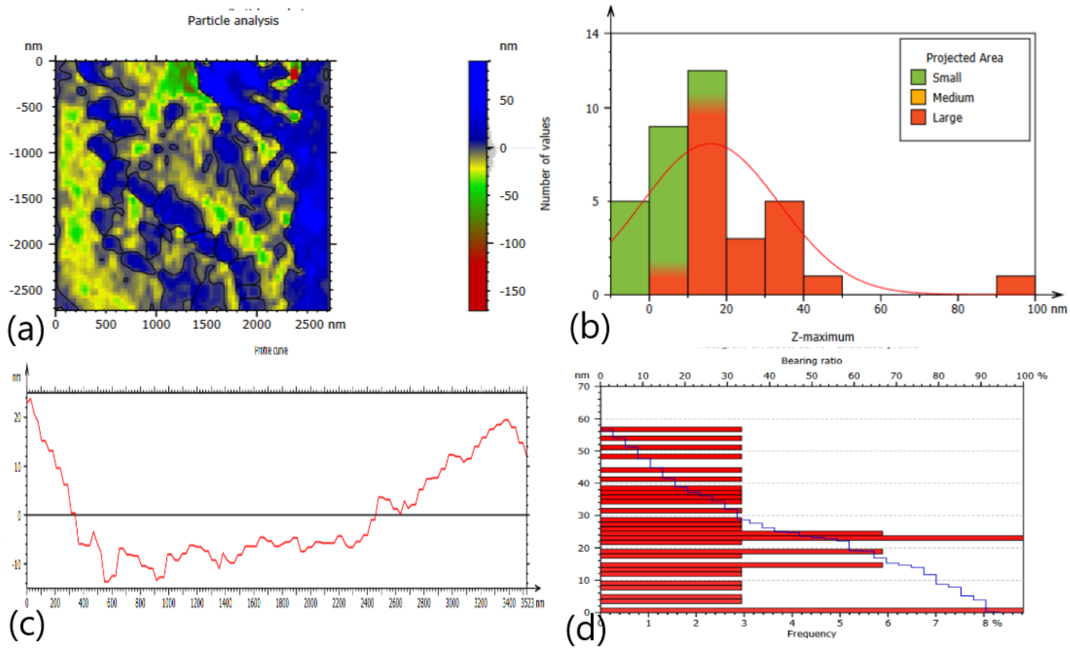


Figure 5.6. AFM analysis of Ti-15Mo-2In alloy with 20 wt.% of salt,) Particle analysis, b) Particles Threshold, c) Profile curve, and d) Histogram & Abbott curve.

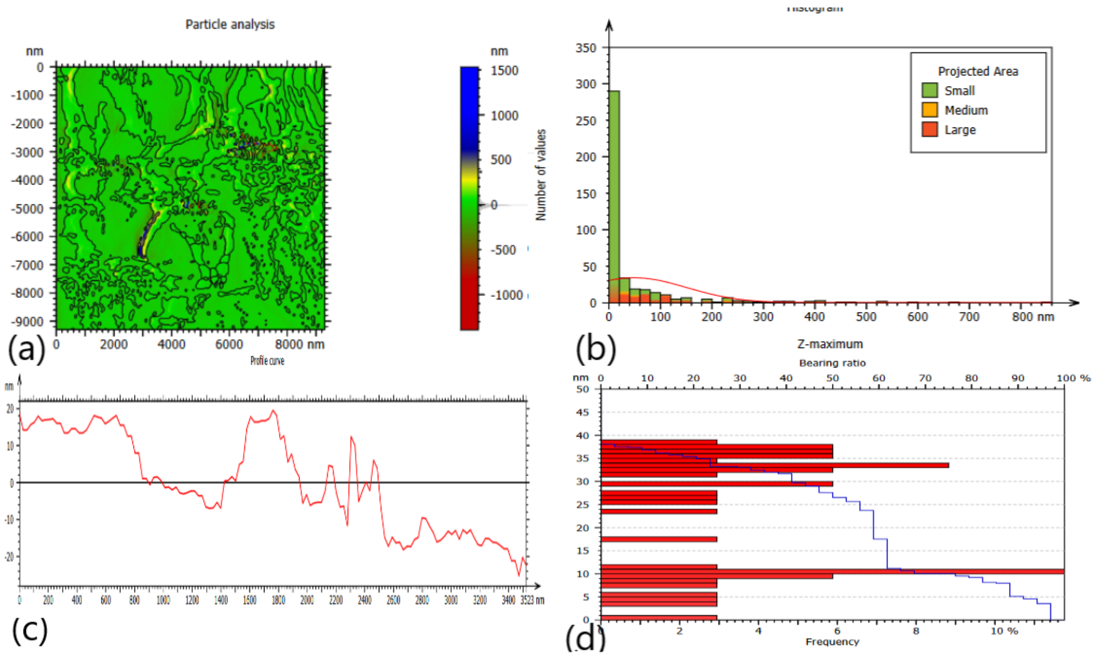


Figure 5.7. AFM analysis of Ti-15Mo-2In alloy with 30 wt.% of salt,) Particle analysis, b) Particles Threshold, c) Profile curve, and d) Histogram & Abbott curve.

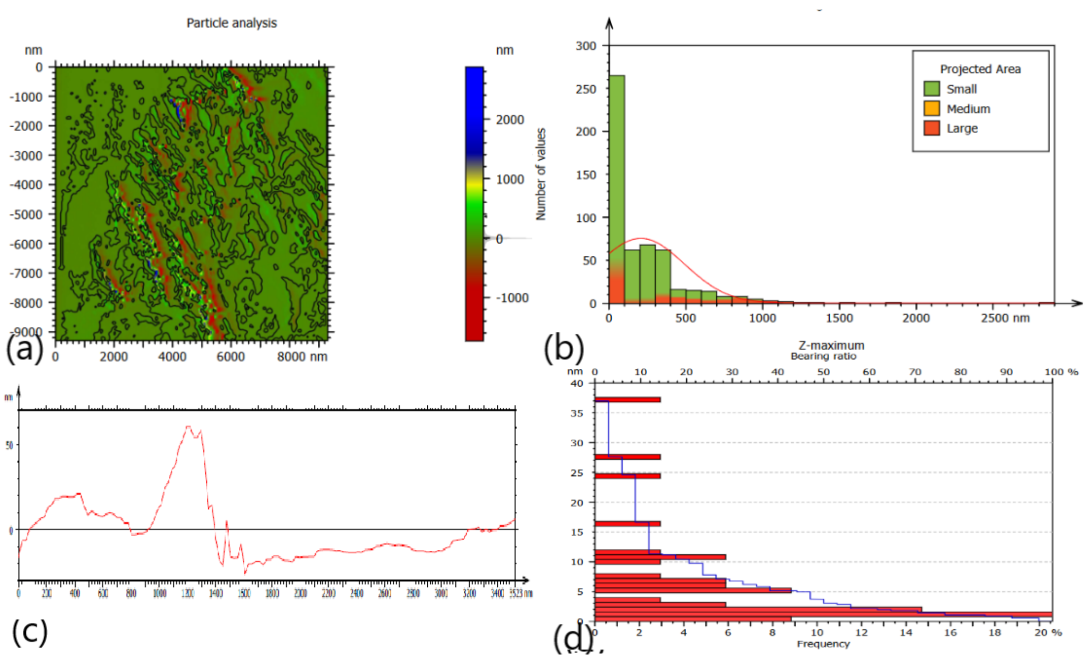


Figure 5.8. AFM analysis of Ti-15Mo-2In alloy with 50 wt.% of salt,) Particle analysis, b) Particles Threshold, c) Profile curve, and d) Histogram & Abbott curve

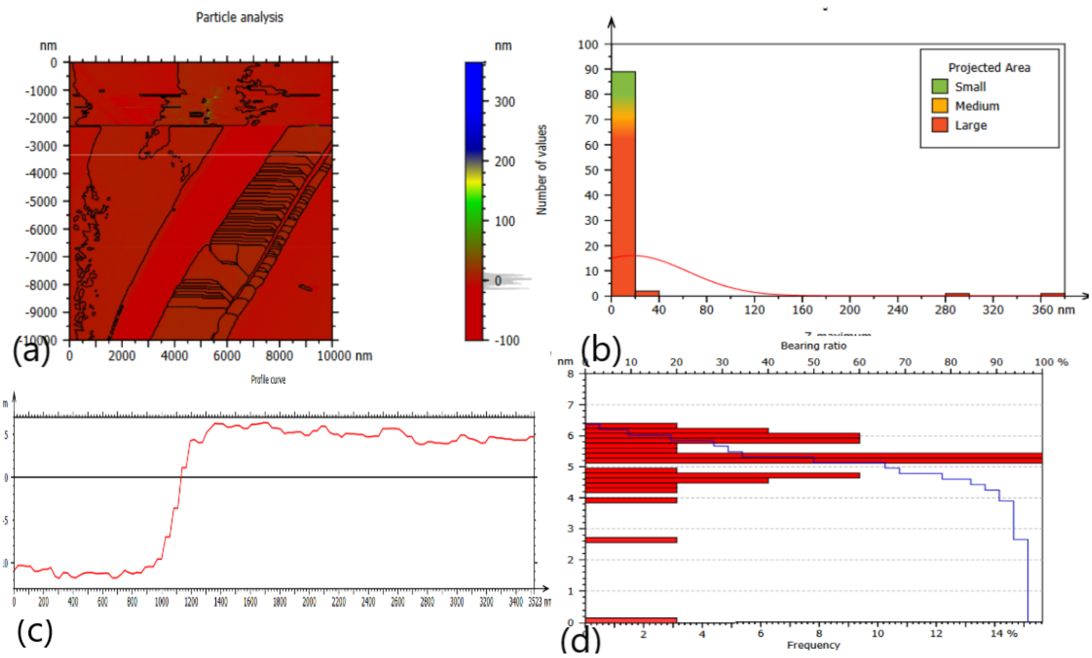


Figure 5.9. AFM analysis of Ti-15Mo-2In alloy with 60 wt.% of salt,) Particle analysis, b) Particles Threshold, c) Profile curve, and d) Histogram & Abbott curve.

Table 5.1. AFM data of prepared alloys.

Alloys	AFM Data			
	Root-mean-square height (Sq) nm	Maximum height (Sz) nm	Arithmetic mean height (Sa) nm	Developed interfacial area ratio (Sdr) %
Ti-15Mo-2In	55.86	1458	23.55	34.31
Ti-15Mo-2In with 20wt.% salt	35.65	926.7	20.54	8.916
Ti-15Mo-2In with 30wt.% salt	34.21	615.3	21.64	9.123
Ti-15Mo-2In with 50wt.% salt	58.63	1452	32.33	40.38
Ti-15Mo-2In with 60wt.% salt	8.392	581.0	3.115	1.307

5.3. PHYSICAL PROPERTIES

The decomposition of the spacing agent produces, among others, gaseous substances, which are responsible for the formation of pores, and formed during the exit of gases[103]. The mixture of powdered titanium with the gas resulting from the decomposition of the solid spreading agent produces: small pores, which appear between the powder particles, also called primary pores, and large pores due to the presence of the gas, called secondary pores. During sintering, the primary pores disappear due to mass transport processes, leaving only the secondary pores. As the latter maintains the shape of the gas bubbles, the shape and a volumetric fraction of the pores in this process are well controlled, obtaining greater sphericity in the present work; the solid spacers are ammonium bicarbonate $(NH_4)HCO_3$ was used, which decompose, at temperatures below 200 ºC, into gaseous components without reacting with titanium. The decomposition of ammonium bicarbonate in the air is clean, producing ammonia, carbon dioxide and water, which are easily eliminated during the process. Wen et al.[104] used ammonium bicarbonate dispersant particles mixed with titanium powder to produce compacts by uniaxial pressing at 100 MPa. The green compacts obtained by this technique were then thermally treated, reaching up to 78% porosities. The structure obtained presented isolated micropores distributed on the walls of interconnected macropores. The authors attributed the formation of these micropores to the volume contraction that occurs during the sintering process. The sponges also showed good mechanical properties expressed in compressive stress values of 35 MPa and Young's modulus of 5.3 GPa. Although this technique produces compacts with high porosity, it does not control the shape of the pores, depending on the homogenization of the powder's salt and titanium mixture.

Adding ammonium carbonate (NH_4HCO_3) resulted in the formation of holes, which led to a reduction in density (as shown in Table 5.2). However, the density drop was not constant and varied across salt concentrations. The injected ammonium carbonate significantly impacts the resulting porosity, with 60% ammonium carbonate yielding the most permeable material. Reducing the particle density generates porosity, which speeds up cell growth after implantation. This scenario shows that the shape and size of the powder components employed in the space holder may affect the pore shapes; in this case, the ammonium carbonate (NH_4HCO_3) space holder material had a comparable porosity to the rectangular pores [103].

Table 5.2. Physical properties of the prepared alloy.

Alloy	Porosity (%)	Density (g/cm^3)
Ti-15Mo-2In	15.41	4.1870
Ti-15Mo-2In with 20wt.% salt	26.27	3.5535
Ti-15Mo-2In with 30wt.% salt	35.26	3.0752
Ti-15Mo-2In with 50wt.% salt	45.68	2.5256
Ti-15Mo-2In with 60wt.% salt	53.83	2.0867

5.4. MECHANICAL PROPERTIES RESULTS

5.4.1. Ultrasonic Wave Test

The mechanical response of biomaterials is best understood by studies of elastic deformation behaviour [123]. The rigidity of a material may be determined by testing its elastic modulus using ultrasonic waves. As illustrated in Figure 5.10 and Table 5.3, the elastic characteristics may be calculated by plugging in the longitudinal speed and shear speed values from Table 5.3 into equations (4.3) and (4.4).

The elastic modulus decreases with increasing salt concentration, as seen in Figure 5.10. When titanium is implanted into the body in its pure form, its elastic modulus is roughly 110 GPa, which is over the ideal threshold and may produce stress shielding. In studies, the elasticity modulus of β-type alloys is between 55 and 124 GPa, which is in accordance with the results obtained for the Ti-15Mo in this study which is lower than that of pure titanium and higher than that of human bone (30GPa)[1]. It was observed through the results that the increase in the percentage of salt led to a decrease in the elastic modulus from 107.462 to 31.492 GPa, as shown in Table 3. As a result, the alloy may be used as a bone replacement, bone strengthening material, prosthetic joint, or implant material.

Table 5.3. Ultrasonic Wave test result of the prepared alloys.

Alloy	Tl (s)	Ts (s)	Longitudinal velocity (m/sec)	Transverse velocity (m/sec)	Poisson's ratio	Elastic modulus (Gpa)
Ti-15Mo-2In	10.4	20.3	0.192	0.098	0.3221	107.462 ± 5.37
Ti-15Mo-2In with 20Wt.% salt	10.7	21	0.186	0.095	0.3225	85.249 ± 4.26
Ti-15Mo-2In with 30Wt.% salt	11.3	22.1	0.176	0.090	0.3230	66.642 ± 3.33
Ti-15Mo-2In with 50Wt.% salt	12.3	24.1	0.162	0.082	0.3239	46.056 ± 2.30
Ti-15Mo-2In with 60Wt.% salt	13.5	26.5	0.148	0.075	0.3248	31.492 ± 1.57

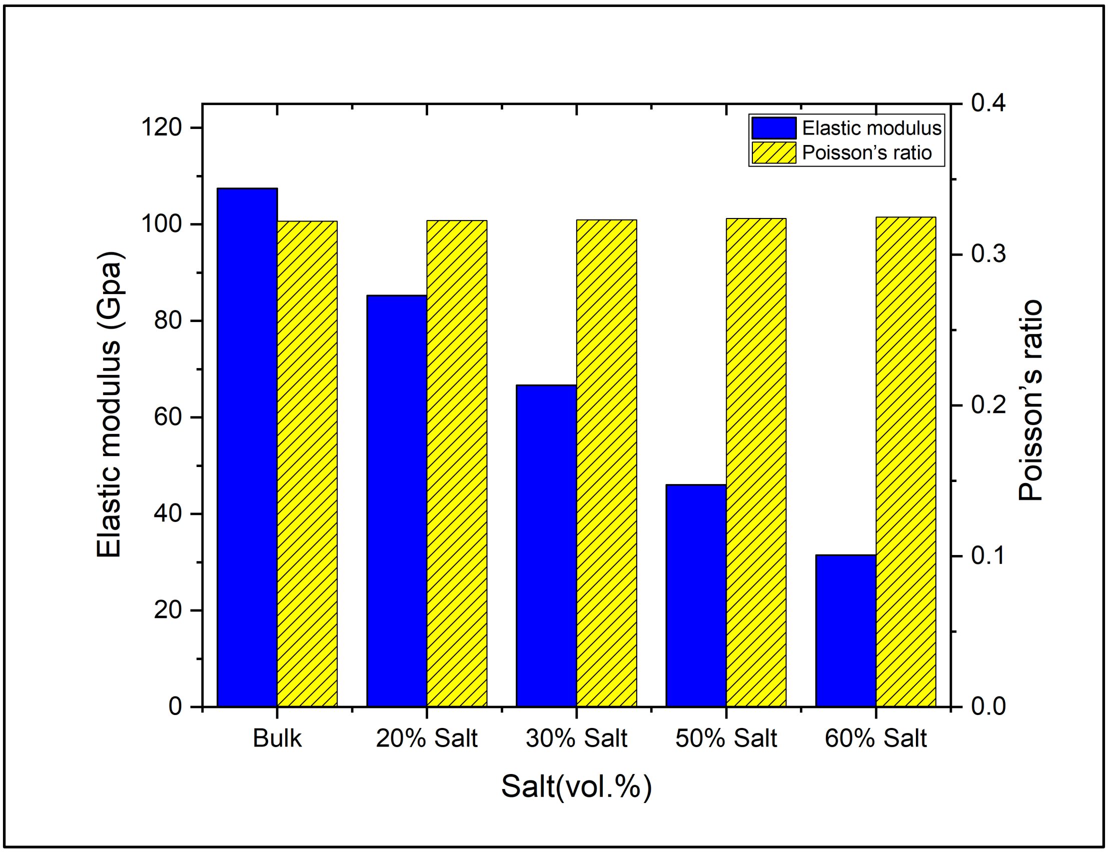


Figure 5.10. Elastic properties of the prepared alloys.

5.4.2. Compression Strength Test

Compression tests demonstrated that the mechanical characteristics changed when ammonium carbonate was added. Figure 5.11 displays the varied compression curves obtained for Ti-15Mo-2In alloy with and without ammonium carbonate at four different percentages (20, 30, 50, and 60 vol.%).

Table 5.4 and Figure 5.12 show that the compressive strength gradually decreases as the salt content rises to 60Vol.%. This behaviour is expected as porosity reduces the compressive strength, as porosity is negatively related to the material's resistance to compression.

The porosity of metals significantly affects their mechanical characteristics [105]. Ti-15Mo-2In alloys have various mechanical characteristics that may be produced by altering the porosity, including yield strength in the 398.88 - 5.356 MPa range, elastic modulus in the 107.46-31.49 GPa and compression strength in the 612.94 - 8.02 MPa range. It is generally agreed upon that implants 'stress shielding' effects may be mitigated if the mechanical characteristics of the implant materials are similar to those of the living bone tissue at the implantation site [43]. Increased porosity and pore size in cancellous bone results in much lower strength and Young's modulus than in cortical bone (110-150 MPa and 18-22 GPa, respectively). Ti-15Mo-2In is a promising new orthopedic implant alloy because of its desirable characteristics and ease of manufacture at low cost. Figure 5.12 shows the correlation between mechanical characteristics and salt concentration. Through the figure 5.12 and table 5.4, it is shown that the increase in the percentage of salt by 50 and 60 %Vol. led to a significant decrease in the values of the mechanical properties, which are considered weak values due to increase the amount of porosity and pore volume .

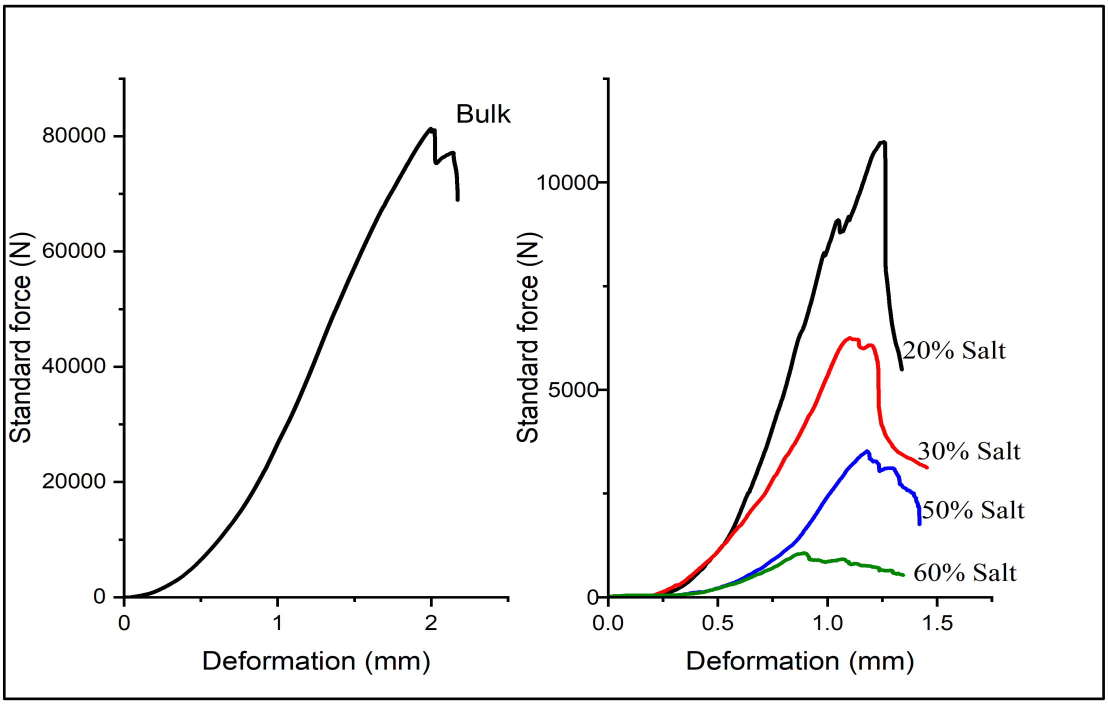


Figure 5.11. Stress-strain curve for the prepared alloys.

Table 5.4. Average mechanical values of the samples.

Sample	**Mechanical Properties**	
	σ_{max} (MPa)	σ_y (MPa)
Ti-15Mo-2In	612.942 ± 30.64	398.886 ± 19.90
Ti-15Mo-2In with 20Vol.% salt	82.710 ± 4.13	46.448 ± 2.32
Ti-15Mo-2In with 30Vol.% salt	47.123 ± 2.35	32.827 ± 1.64
Ti-15Mo-2In with 50Vol.% salt	26.565 ± 1.32	9.591 ± 0.47
Ti-15Mo-2In with 60Vol.% salt	8.029 ± 0.40	5.356 ± 0.26

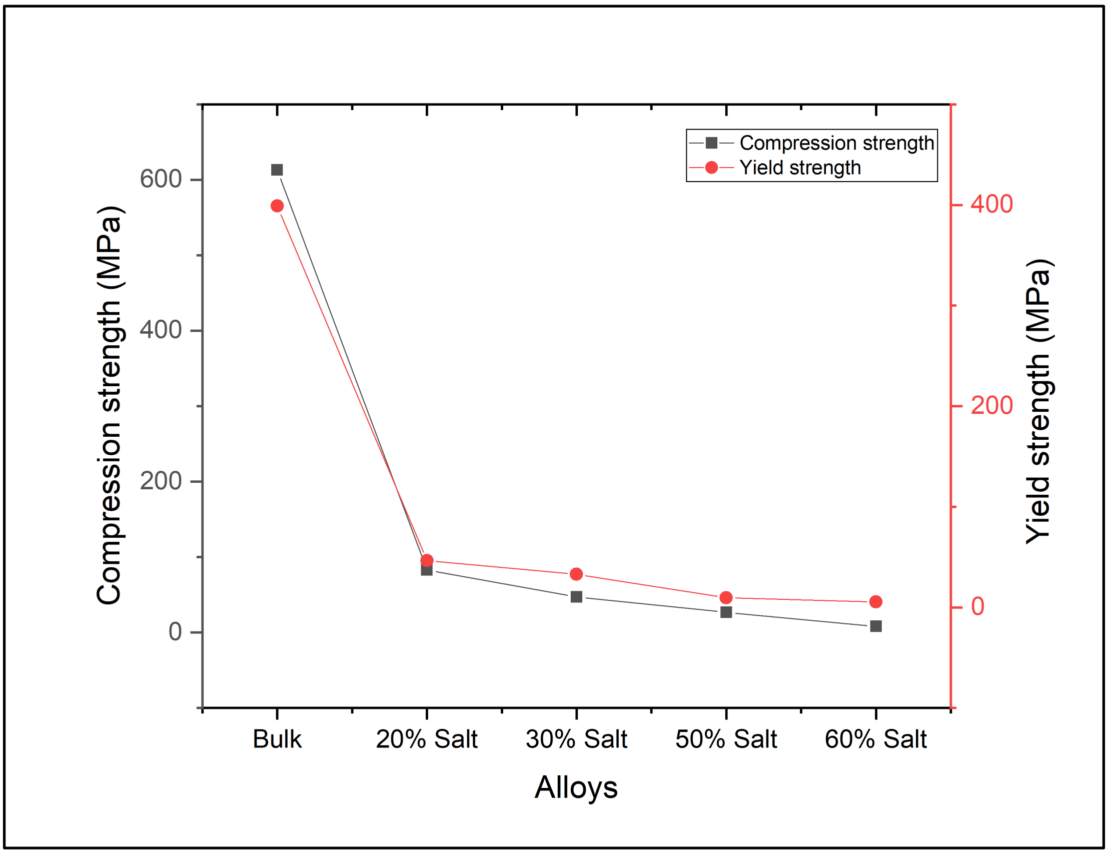


Figure 5.12. Mechanical properties of tested alloys.

5.4.3. Hardness Test

Figure 5.13 displays the results of Brinell hardness tests performed on Ti-15Mo-2In alloy specimens exposed to a range of salt concentrations. Results show that hardness values in the Ti-15Mo-2In systems vary with salt concentration. An increase in salt content from 20% to 60% led to a decrease in the brinel hardness (HB) values by about 40%. The effect of pores on the mechanical properties of produced Ti-15Mo-2In was investigated by evaluating hardness and salt content. Figure 5.13 displays the produced material hardness as a function of porosity for the deposition conditions of this study. In agreement with Karthikeyan et al. the results show that the hardness of the material falls as the percentage of porosity increases. Microhardness increases with increasing density and decreasing porosity [106],

The porosity of a material can have a significant impact on its mechanical properties, including its hardness. In the case of titanium alloy, a commonly used material in various industries due to its excellent strength-to-weight ratio, pores, or voids can affect its hardness in several ways.

Firstly, porosity can reduce the density of the material, which in turn can lower its hardness. Hardness is often defined as a material's resistance to deformation or penetration, and a lower-density material will typically be more easily penetrated or deformed.

Secondly, pores or voids can also create stress concentration points within the material, leading to localized deformation and cracking. This can further reduce the material's hardness and overall mechanical strength.

Also, porosity can affect the material's microstructure, impacting its hardness. For example, if the pores or voids are not properly filled during the production process, they can create areas of lower density or weaker bonding between the titanium alloy grains, reducing their hardness.

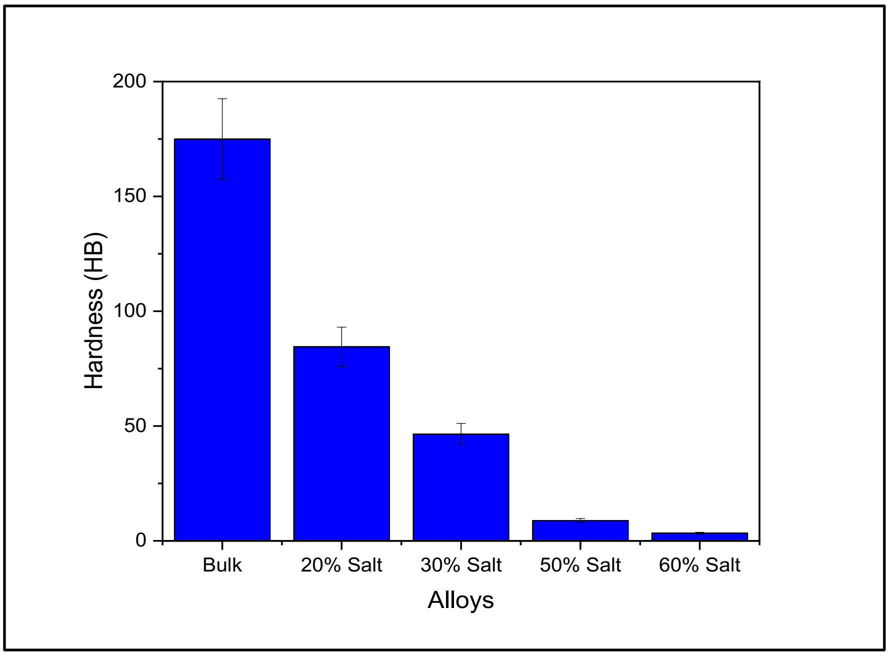


Figure 5.13. Brinell hardness of alloys .

5.5. ANTIBACTERIAL ANALYSIS

In this work, the Ti-15Mo-2In with a standard antibiotic disc of ampicillin was utilized to conduct an antibacterial test against E.coli and S.aureus using the disk diffusion technique to understand the variations in antibacterial capabilities. Within 24 hours, an inhibitory zone formed around the disks holding the samples in the agar plates. According to statistical analysis, the inhibitory zone width surrounding the Ti-15Mo-2In samples vary considerably against the same bacterium. As shown in Figure 5.14 and Table 5.5, the widths of the inhibition zones generated around the Ti-15Mo-2In samples were 15 mm against the S.aureus, compared to 29 and 14 mm for a typical antibiotic disc of Cefoxitin and Penicillin respectively, while there no effect against E.Coils compared to 30 and 13 mm for a typical antibiotic disc of Cefoxitin and Penicillin respectively. The result indicates good efficiency against S.aureus bacteria. Titanium alloys have been found to have antibacterial properties against various microorganisms, including Staphylococcus aureus (S. aureus). It is believed that the antibacterial properties of Ti alloys are due to the release of metal ions from the implant's surface, which can inhibit bacterial growth. In particular, research has shown that Ti6Al4V alloy, commonly used in orthopedic and dental implants, has significant antibacterial activity against S. aureus. One study found that Ti6Al4V alloy inhibited the growth of S. aureus by up to 99.9% after 24 hours of exposure. The exact mechanism by which Ti alloys exert their antibacterial effect is not fully understood. However, it is believed to involve the release of metal ions such as titanium, aluminum, and vanadium, which can disrupt bacterial cell membranes and inhibit bacterial metabolism.

There are several possible reasons why a Ti alloy may not produce an inhibition zone against Escherichia coli (E. coli) bacteria. Resistance of E. coli to the antibacterial properties of Ti alloy: E. coli is known to be resistant to certain metals, such as silver, which are commonly used as antibacterial agents. E. coli may also resist Ti alloy's antibacterial properties, which could explain why no inhibition zone is observed. Different bacterial growth requirements. E. coli and Staphylococcus aureus (S. aureus) have different growth requirements and may respond differently to the same antibacterial agent. For example, S. aureus is a facultative anaerobe, meaning it can grow in the presence or absence of oxygen, while E. coli is an obligate anaerobe, requiring an oxygen-free environment to grow. It is possible that the growth requirements of E. coli are not compatible with the antibacterial properties of Ti alloy. Experimental conditions. The absence of an inhibition zone could be due to experimental conditions, such as the concentration of Ti alloy or the length of exposure time. It is possible that different experimental conditions could produce different results, including the presence or absence of an inhibition zone.

Table 5.5. Inhibition Zone Diameters

S.Aureus 25923		**E.Coli 25922**	
Test Material	Zone Diameter	Test Material	Zone Diameter
Ti-15Mo-2In	15 mm	Ti-15Mo-2In	-
Control(Cefoxitin 30 µg)	29 mm	Control(Cefoxitin 30 µg)	30 mm
Control (Penicillin G 10 µg)	14 mm	Control (Penicillin G 10 µg)	13 mm

Figure 5.14. Inhibition zone of the S.Aureus (25923) and E.Coli (25922) inoculated bacteria plate with the Ti-15Mo-2In sample.

PART 6

CONCLUSIONS AND RECOMMENDATION

6.1 CONCLUSIONS

Ti6Al4V alloy, widely used in biomaterials, causes many diseases, such as Alzheimer's, due to its high elastic modulus and toxic properties. This necessitated the production of new-generation titanium alloys. The results obtained within the scope of this thesis study, which is based on the development, production and characterization of new-generation titanium alloys, which are considered to be used as biomaterials, are promising. The results obtained are summarized below:

i. The production of porous new-generation biocompatible triple titanium alloys from Ti, Mo, In and different concentrations of salt elements has been successfully accomplished by powder metallurgy methods.
ii. When the XRD results were examined, α – Ti, β – Ti and Ti_3In_4 phases were observed in all the alloys produced, and it was seen that the alloying elements added as additives formed a solid solution without forming a new phase. Adding Mo to the structure ensured the β – Ti phase formation since they are β stabilizers.
iii. SEM and EDS results show that the alloys produced have a homogeneous compositional distribution. Spectrum analyses taken from different parts of the alloys showed that the alloy compositions were in the targeted compositions.
iv. The percentage of porosity increases with the increase in the salt content. It showed that when the salt concentration reached 60 %, the porosity was 53.83%, while the density decreased from 4.18 to 2.08 g/cm^3.
v. The values of modulus of elasticity decrease with an increase in the addition of salt concentration compared with the Ti-15Mo-2In alloy without additive due to an increase in the porosity of the final structure of the prepared alloy.

vi. The compressive strength and hardness decrease with an increase in the porosity; the presence of porosity in a titanium alloy can have a negative impact on its hardness by reducing its density, creating stress concentration points, and affecting its microstructure. It is, therefore, essential to optimize the porosity during production to ensure the material has the desired mechanical properties.

vii. Ti alloys have been shown to have promising antibacterial properties against S. aureus, which could make them a valuable material for use in medical implants and other applications where bacterial colonization is a concern.

viii. The absence of an inhibition zone for Ti alloy against E. coli could be due to various factors, including bacterial resistance, different growth requirements, or experimental conditions.

6.2 RECOMMENDATIONS

Porous Ti-15Mo-2In alloy is a promising material for orthopedic and dental implants due to its excellent biocompatibility, mechanical properties, and ability to promote osseointegration. Here are some recommendations for preparing porous Ti-15Mo-2In alloy:

1. Fabrication method: Porous Ti-15Mo-2In alloy can be prepared using various techniques, including powder metallurgy, additive manufacturing, and foaming. The choice of fabrication method will depend on factors such as the desired porosity, pore size, and material's mechanical properties.
2. Porosity and pore size: The porosity and pore size of the porous Ti-15Mo-2In alloy can be controlled by adjusting the processing parameters, such as the particle size of the starting material, the sintering temperature and time, and the foaming agent concentration. It is important to optimize these parameters to achieve the desired mechanical properties and biological performance of the material.
3. Surface modification: The surface of the porous Ti-15Mo-2In alloy can be modified to enhance its biological performance, such as by coating it with bioactive materials or incorporating growth factors or drugs into the pores. This can improve the osseointegration of the implant and reduce the risk of infection.
4. Mechanical testing: The mechanical properties of the porous Ti-15Mo-2In alloy, such as compressive strength, elastic modulus, and fatigue resistance, should be tested to ensure that they meet the requirements for orthopedic and dental applications. Testing should be performed both in vitro and in vivo to evaluate the long-term stability and performance of the material.

Overall, preparing porous Ti-15Mo-2In alloy requires careful consideration of the processing parameters, surface modification, and mechanical testing to ensure the material meets orthopedic and dental implant application requirements.

REFERENCES

1. Al-Humairi, A.N.S., et al., BIOMATERIALS: Multidisciplinary approaches and their related applications,***White Falcon Publishing*** (2020).
2. Chen, W., et al., "In vitro anti-bacterial and biological properties of magnetron co-sputtered silver-containing hydroxyapatite coating". ***Biomaterials***, 27(32):5512-5517 (2006).
3. Losina, E., et al., "Early failures of total hip replacement: effect of surgeon volume". ***Arthritis Rheumatism: Official Journal of the American College of Rheumatology,*** 50(4):1338-1343 (2004).
4. Harris, L.G., et al., "Staphylococcus aureus adhesion to standard micro-rough and electropolished implant materials". ***Journal of Materials Science: Materials in Medicine,*** 18(6):1151-1156 (2007).
5. Biehl, V. and J.J. Breme, "Metallic biomaterials". ***Materialwissenschaft und Werkstofftechnik***, 32(2):137-141 (2001).
6. Niinomi, M., T. Hattori, and S. Niwa, "Material characteristics and biocompatibility of low rigidity titanium alloys for biomedical applications". ***Biomaterials in orthopedics***, 22:41 (2004).
7. Oshida, Y., et al., "Fractal dimension analysis of mandibular bones: toward a morphological compatibility of implants". ***Bio-medical materials engineering,*** 4(5):397-407 (1994).
8. Long, M. and H. Rack, "Titanium alloys in total joint replacement—a materials science perspective". ***Biomaterials***, 19(18):1621-1639 (1998).
9. Harrysson, O.L. and D.R. Cormier, "Direct fabrication of custom orthopedic implants using electron beam melting technology". **A*dvanced Manufacturing Technology for Medical Applications: Reverse Engineering, Software Conversion Rapid Prototyping*:**191-206 (2005).

10. Niinomi, M., "Mechanical properties of biomedical titanium alloys". ***Materials Science Engineering: A***, 243(1-2):231-236 (1998).
11. Williams, J. and G. Luetjering, "The effect of slip length and slip character on the properties of titanium alloys". ***Titanium***, 80:671-681 (1980).
12. Froes, F., et al., "The technologies of titanium powder metallurgy". ***Jom***, 56:46-48 (2004).
13. Ding, R., Z. Guo, and A.J. Wilson, "Microstructural evolution of a Ti–6Al–4V alloy during thermomechanical processing". ***Materials Science Engineering: A***, 327(2):233-245 (2002).
14. Oshida, Y. Requirements for successful biofunctional implants. in The 2nd ***Symposium International of Advanced Bio-Materials***. (2000).
15. Bagno, A. and C.J. Di Bello, "Surface treatments and roughness properties of Ti-based biomaterials". ***Journal of materials science: materials in medicine***, 15(9):935-949 (2004).
16. Liu, X., P.K. Chu, and C.J. Ding, "Surface modification of titanium, titanium alloys, and related materials for biomedical applications". ***Materials Science Engineering: R: Reports,*** 47(3-4):49-121 (2004).
17. Massaro, C., et al., "Comparative investigation of the surface properties of commercial titanium dental implants. Part I: chemical composition". ***Journal of Materials Science: Materials in Medicine***, 13(6):535 (2002).
18. Popat, K.C., et al., "Decreased Staphylococcus epidermis adhesion and increased osteoblast functionality on antibiotic-loaded titania nanotubes". ***Biomaterials***, 28(32):4880-4888 (2007).
19. Yamada, Y., et al. Preparation of bioactive porous titanium-molybdenum alloy through powder metallurgy. in Materials Science Forum. 2009. Trans Tech Publ.
20. Kolobov, Y., M. Ivanov, and V.J. Itin, "Porous super-elastic Ti-Mo-based alloys for osteosynthesis". ***Advanced Materials Processes***, 163(7):68-68 (2005).
21. Zhou, Y.-L. and D.-M.J. Luo, "Microstructures and mechanical properties of Ti–Mo alloys cold-rolled and heat treated". ***Materials Characterization***, 62(10):931-937 (2011).

22. Xu, W., et al., "Mechanical properties, in vitro corrosion resistance and biocompatibility of metal injection molded Ti-12Mo alloy for dental applications". J***ournal of the Mechanical Behavior of Biomedical Materials,*** 88:534-547 (2018).
23. Lee, D., et al., "Microstructures and mechanical properties of Ti and Mo micro-alloyed medium Mn steel". ***Materials Science Engineering: A***, 706:1-14 (2017).
24. Xu, J., et al., "Effects of Mo contents on the microstructure, properties and cytocompatibility of the microwave sintered porous Ti-Mo alloys". ***Materials Science Engineering: C***, 97:156-165 (2019).
25. Calin, M., et al., "Elastic softening of β-type Ti–Nb alloys by indium (In) additions". ***Journal of the mechanical behavior of biomedical materials***, 39:162-174 (2014).
26. Du, Z., et al., "The influences of process annealing temperature on microstructure and mechanical properties of near β high strength titanium alloy sheet". ***Materials,*** 12(9):1478 (2019).
27. Li, P., et al., "Microstructural and mechanical properties of β-type Ti–Mo–Nb biomedical alloys with low elastic modulus". ***Journal of Alloys Compounds,*** 815:152412 (2020).
28. Al-Murshdy, J.M.S., Al-Deen, H. H. J., & Hussein, S. R, "Investigation of the Effect of Indium Addition on the Mechanical and Electrochemical Properties of the Ti–15Mo Biomedical Alloy". ***Journal of Bio-and Tribo-Corrosion,*** 7(4):1-18 (2021).
29. Chen, Q. and G.A.J. Thouas, "Metallic implant biomaterials". ***Materials Science Engineering: R: Reports,*** 87:1-57 (2015).
30. Carnicer-Lombarte, A., et al., "Foreign body reaction to implanted biomaterials and its impact in nerve neuroprosthetics". ***Frontiers in Bioengineering Biotechnology***:271 (2021).
31. Todros, S., M. Todesco, and A.J. Bagno, "Biomaterials and their biomedical applications: From replacement to regeneration". ***Processes***, 9(11):1949 (2021).
32. Zafar, M.S., et al., "Customized therapeutic surface coatings for dental implants". ***Coatings***, 10(6):568 (2020).

33. Bergmann, G., et al., "Hip contact forces and gait patterns from routine activities". ***Journal of biom*echanics**, 34(7):859-871 (2001).
34. Dos Santos, V., R.N. Brandalise, and M. Savaris, Engineering of biomaterials. Vol. 10. 2017: Springer.
35. Park, J. and R.S. Lakes, Biomaterials: an introduction. 2007: Springer Science & Business Media.
36. Prasad, K., et al., "Metallic biomaterials: current challenges and opportunities". ***Materials***, 10(8):884 (2017).
37. GÜLTEKİN, M.Z. and Y.M. DİNÇEL, "Biomaterials Used in Orthopedics and Their Properties". **A*DVANCES IN HEALTH SCIENCES***:231.
38. Wilson, J.J., "Metallic biomaterials: State of the art and new challenges". ***Fundamental Biomaterials: Metals***:1-33 (2018).
39. Liu, X., et al., "Binary titanium alloys as dental implant materials—a review". ***Regenerative biomaterials***, 4(5):315-323 (2017).
40. Hussein, M.A., A.S. Mohammed, and N. Al-Aqeeli, "Wear characteristics of metallic biomaterials: a review". ***Materials***, 8(5):2749-2768 (2015).
41. Ratner, B.D., et al., Biomaterials science: an introduction to materials in medicine. 2004: Elsevier.
42. GÜVEN, Ş.J., "Biyouyumluluk ve biyomalzemelerin seçimi". ***Mühendislik Bilimleri ve Tasarım Dergisi***, 2(3):303-311 (2014).
43. Geetha, M., et al., "Ti based biomaterials, the ultimate choice for orthopaedic implants–a review". ***Progress in materials science***, 54(3):397-425 (2009).
44. Davis, R., et al., "A comprehensive review on metallic implant biomaterials and their subtractive manufacturing". **T*he International Journal of Advanced Manufacturing Technology***, 120(3-4):1473-1530 (2022).
45. de Viteri, V.S. and E.J. Fuentes, "Titanium and titanium alloys as biomaterials". ***Tribology-fundamentals advancements***, 1(5):154-181 (2013).
46. Davoodi, E., et al., "Additively manufactured metallic biomaterials". **B*ioactive Materials***, 15:214-249 (2022).
47. Moghadasi, K., et al., "A review on biomedical implant materials and the effect of friction stir based techniques on their mechanical and tribological properties". ***Journal of Materials Research Technology***, (2022).

48. Kamachimudali, U., T. Sridhar, and B.J. Raj, "Corrosion of bio implants". ***Sadhana***, 28:601-637 (2003).
49. Quinn, J., et al., "Titanium for orthopedic applications: an overview of surface modification to improve biocompatibility and prevent bacterial biofilm formation". ***IScience***, 23(11):101745 (2020).
50. Dill, H.G.J., "The "chessboard" classification scheme of mineral deposits: Mineralogy and geology from aluminum to zirconium". ***Earth-Science Reviews,*** 100(1-4):1-420 (2010).
51. Lütjering, G. and J.C. Williams, "Special properties and applications of titanium". ***Titanium***:383-415 (2007).
52. Boettcher, C., T. Bell, and H.J. Dong, "Surface engineering of timet 550 with oxygen to form a rutile-based, wear-resistant coating". ***Metallurgical Materials Transactions A,*** 33:1201-1211 (2002).
53. Stachowiak, G.W. and A.W. Batchelor, Engineering tribology. 2013: Butterworth-heinemann.
54. Zeng, L. and T.J. Bieler, "Effects of working, heat treatment, and aging on microstructural evolution and crystallographic texture of α, α′, α ″and β phases in Ti–6Al–4V wire". ***Materials Science Engineering: A***, 392(1-2):403-414 (2005).
55. Leyens, C. and M. Peters, Titanium and titanium alloys: fundamentals and applications: ***Wiley Online Library***, (2006).
56. Froes, F. and M. Whittaker, Titanium and its alloys. 2022, Elsevier.
57. Kolli, R.P. and A.J. Devaraj, "A review of metastable beta titanium alloys". **Metals**, 8(7):506 (2018).
58. Eisenbarth, E., et al., "Biocompatibility of β-stabilizing elements of titanium alloys". ***Biomaterials***, 25(26):5705-5713 (2004).
59. Zhang, L.-C., L.-Y. Chen, and L. Wang, "Surface modification of titanium and titanium alloys: technologies, developments, and future interests". ***Advanced Engineering Materials,*** 22(5):1901258 (2020).
60. Hoque, M.E., et al., "Titanium and titanium alloys in dentistry: current trends, recent developments, and future prospects". ***Heliyon***:e11300 (2022).
61. Eliaz, N.J., "Corrosion of metallic biomaterials: a review". ***Materials***, 12(3):407 (2019).

62. Martins Júnior, J.R.S., et al., "Preparation and characterization of Ti-15Mo alloy used as biomaterial". ***Materials Research***, 14:107-112 (2011).
63. Black, J. and G. Hastings, Handbook of biomaterial properties. 2013: ***Springer*** Science & Business Media.
64. Marquardt, B. and Shetty, R., "Beta Titanium Alloy Processed for High Strength Orthopedic Applications", 2 (9): (2005).
65. Ho, W., C. Ju, and J.C.J. Lin, "Structure and properties of cast binary Ti–Mo alloys". ***Biomaterials***, 20(22):2115-2122 (1999).
66. Zhang, W.-d., et al., "Elastic modulus of phases in Ti–Mo alloys". ***Materials Characterization,*** 106:302-307 (2015).
67. Delvat, E., et al., "Microstructure, mechanical properties and cytocompatibility of stable beta Ti–Mo–Ta sintered alloys". **J*ournal of the mechanical behavior of biomedical materials,*** 1(4):345-351 (2008).
68. Xu, Y. and H.J. Nomura, "Homogenizing analysis for sintering of bio-titanium alloy (Ti-5Al-2.5 Fe) in MIM process". ***Journal of the Japan Society of Powder Powder Metallurgy,*** 48(11):1089-1096 (2001).
69. Antunes, R.A. and M.C.L. de Oliveira, "Corrosion fatigue of biomedical metallic alloys: mechanisms and mitigation". ***Acta biomaterialia***, 8(3):937-962 (2012).
70. Bania, P.J., "Beta titanium alloys and their role in the titanium industry". **Jom**, 46:16-19 (1994).
71. Zardiackas, L.D., D.W. Mitchell, and J.A.J. Disegi, "Characterization of ti-15Mo beta titanium alloy for orthopaedic". ***Medical applications of titanium its alloys: the material biological,*** (1272):60 (1996).
72. Espevik, S., G. Osilo, and A. Lodding, "Oxidation of noble metal alloys for porcelain veneer crowns". ***Acta Odontologica Scandinavica***, 37(6):323-328 (1979).
73. Morris, H., et al., "Casting alloys: the materials and" the clinical effects"". ***Advances in Dental research,*** 6(1):28-31 (1992).
74. Kim, T.-I., et al., "New titanium alloys for biomaterials: a study of mechanical and corrosion properties and cytotoxicity". **B*io-medical materials engineering,*** 7(4):253-263 (1997).

75. Wang, Q., et al., "Development and properties of Ti–In binary alloys as dental biomaterials". ***Materials Science Engineering: C***, 33(3):1601-1606 (2013).
76. Han, M.-K., et al., "Effect of indium content on the microstructure, mechanical properties and corrosion behavior of titanium alloys". ***Metals***, 5(2):850-862 (2015).
77. Nakai, M., M. Niinomi, and T. Oneda, "Improvement in fatigue strength of biomedical β-type Ti–Nb–Ta–Zr alloy while maintaining low Young's modulus through optimizing ω-phase precipitation". ***Metallurgical Materials Transactions A***, 43:294-302 (2012).
78. Ng, H.P., et al., "Improving sinterability of Ti–6Al–4V from blended elemental powders through equal channel angular pressing". ***Materials Science Engineering: A***, 565:396-404 (2013).
79. McCracken, C.G., C. Motchenbacher, and D.P. Barbis, "REVIEW OF TITANIUM-POWDER-PRODUCTION METHODS". ***International journal of powder metallurgy***, 46(5) (2010).
80. Mitkov, M. and D. Bozˇić, "Hydride-dehydride conversion of solid Ti6Al4V to powder form". ***Materials characterization,*** 37(2-3):53-60 (1996).
81. Yalçınkaya, F.J., "Toz Metalurji Metodu İle Üretilen Alüminyum Matrisli Ve Alümina Takviyeli Kompozit Malzemenin Abrasiv Aşınma Davranışının İncelenmesiy". ***Gazi Üniversitesi***, (2018).
82. Dondi, M., "Powder Granulation and Compaction", Ed., ***Elsevier***, Oxford, (2021).
83. Totten, G. E., Xie, L.and Funatani, K., "Handbook of Mechanical Alloy Design", 1st Edition. Ed., ***CRC Press***, Boca Raton, (2003).
84. Danninger, H., R. Calderon, and C.J. Gierl-Mayer, "Powder metallurgy and sintered materials". ***Additive Manufacturing,*** 19(4):1-57 (2017).
85. Arifin, A., et al., "Material processing of hydroxyapatite and titanium alloy (HA/Ti) composite as implant materials using powder metallurgy: a review". ***Materials Design***, 55:165-175 (2014).
86. Archbold, J., Sintering temperature effects on the mechanical properties of porous-coated Ti-6Al-4V ELI alloy, in Depatment of metallurgy and material science 2001, university of Toronto.
87. Committee, A.I.H., ASM handbook. Vol. 7. 1998: ASM International.

88. Donachie, M., Titanium a technical guide, the material information society. 2000, USA.
89. Frary, M., et al., "Microstructure and mechanical properties of Ti/W and Ti–6Al–4V/W composites fabricated by powder-metallurgy". ***Materials Science Engineering: A**,* 344(1-2):103-112 (2003).
90. Okazaki, Y. and K. Chinzei, Development of Orthopedic Implants with Highly Biocompatible Ti Alloys, High Entropy Materials-Microstructures and Properties, ***IntechOpen,*** (2022).
91. Dikici, B., et al., "Corrosion of metallic biomaterials". ***Advances in Metallic Biomaterials: Tissues, Materials Biological Reactions***:275-303 (2015).
92. Manam, N., et al., "Study of corrosion in biocompatible metals for implants: A review". ***Journal of alloys compounds***, 701:698-715 (2017).
93. Severino Martins Jr, J.R. and C.R.J. Grandini, "Structural characterization of Ti-15Mo alloy used as biomaterial by Rietveld method". ***Journal of Applied Physics***, 111(8):083535 (2012).
94. Dąbrowski, R.J., "The kinetics of phase transformations during continuous cooling of the Ti6Al4V alloy from the single-phase ß range". **A*rchives of Metallurgy Materials***, 56(3):703-707 (2011).
95. Bongaerts, J., D. Rossetti, and J.J. Stokes, "The lubricating properties of human whole saliva". ***Tribology Letters**,* 27:277-287 (2007).
96. Dursun, Ö., et al., "Synthesis, characterization and dry sliding wear behavior of in-situ formed TiAl 3 precipitate reinforced A356 alloy produced by mechanical alloying method". ***Materials Research***, 18:813-820 (2015).
97. Hauptmann, A. and A.J. Hauptmann, "Macro-and Microstructure of Metals". ***Archaeometallurgy–Materials Science Aspects***:433-443 (2020).
98. Jayaraman, M., et al., "Influence of titanium surfaces on attachment of osteoblast-like cells in vitro". ***Biomaterials***, 25(4):625-631 (2004).
99. Salleh, F.M., et al., "Effect of space holders on fabrication of porous titanium alloy-hydroxyapatite composite through powder injection molding". ***Sains Malaysiana**,* 46(9):1651-1657 (2017).
100. de Freitas Daudt, N., et al., "Manufacturing of highly porous titanium by metal injection molding in combination with plasma treatment". **J*ournal of Materials Processing Technology**,* 239:202-209 (2017).

101. Vlcak, P., et al., "Surface pre-treatments of Ti-Nb-Zr-Ta beta titanium alloy: The effect of chemical, electrochemical and ion sputter etching on morphology, residual stress, corrosion stability and the MG-63 cell response". ***Results in Physics***, 28:104613 (2021).
102. Likibi, F., et al. Bone integration and apposition of porous and non porous metallic orthopaedic biomaterials. **Annales de chirurgie**, 130(4):235-41 (2005).
103. Laptev, A., et al., "Study of production route for titanium parts combining very high porosity and complex shape". ***Powder metallurgy***, 47(1):85-92 (2004).
104. Wen, C., et al., "Fabrication and characterization of autogenous titanium foams". ***Eur Cells Mater***, 1:61-2 (2001).
105. Wen, C., et al., "Processing of biocompatible porous Ti and Mg". ***Scripta materialia,*** 45(10):1147-1153 (2001).
106. Karthikeyan, J., et al. Cold spray processing of titanium powder. **ASM International ,**(2000).

RESUME

Hussein Basim Baqer OBAIDA is a material engineer who graduated from the Faculty of Engineering, University of AL-MUSTANSİNYA - Iraq. He received His Bachelor’s degree in 2012. He is currently studying for her Master's degree at Karabük University in the field of Materials Engineering.