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THE EFFECT OF VOLTAGE AND TIME ON OXIDE NANOTUBES OF TITANIUM ALLOY BY ANODIZING FOR METAL IMPLANT

By

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DECLARATION

I hereby declare that I have conducted and completed the research work and written the dissertation entitled “Titanium Niobium with Oxide nanotube as metal implant”. I also declare that it has not been previously submitted for award of any degree or diploma or other similar tittle of this for any other examining body or University.

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TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLE

TITANIUM NIOBIUM WITH OXIDE NANOTUBE AS METAL IMPLANT

ABSTRAK

The Ti-Nb alloys have been widely studied due to their high biocompatibility and excellent mechanical properties for biomedical applications. The objective of this study is to study the effect of the anodization time and the enhancement of the bioactivity. TiO_2 nanotubes is achieve by electrochemical anodizing. Nanotubes with different size of diameter and length were grown on Ti-Nb substrates with kept constant 10 V and 30 V at 5 min, 35 min and 65 min via electrochemical anodization. Scanning electron microscopy (SEM) imaging shows that vertically aligned nanotubular structures form on the surface of Ti-Nb alloy substrates. The anodizing voltage and time are influencing the diameter and length of nanotubes. XRD analysis confirmed that there is the formation of TiO_2 layer and β phase titanium. The different of diameter and length of nanotubes can affect the properties of the Ti-Nb alloy such as corrosion resistance and mechanical. Anodizing at 30 V for 35 minutes and 65 minutes results in nanotubes with diameters of 45.53 nm and 50.66 nm, respectively. Anodizing at 30 V for 35 minutes and 65 minutes results in nanotubes with lengths of 1225.77 nm and 2250 nm. Anodizing at 10 volts for 35 minutes and 65 minutes produces nanotubes with diameters of 20.18 nm and 21.64 nm. The length of the nanotube is 106.98 nm for anodizing at 10 V for 35 minutes and 201.13 nm for 65 minutes. The surface of Ti-Nb anodized at 30 V promoted enhance cell growth as the time anodizing is increase. The application of TiO_2 nanotubes is improves cell adhesion which is important in the biomedical application.

TITANIUM NIOBIUM DENGAN OKSIDA NANOTUBE SEBAGAI IMPLAN LOGAM

ABSTRAK

Aloi Ti-Nb telah dikaji secara meluas kerana biokompatibiliti tinggi dan sifat mekanikal yang sangat baik untuk aplikasi bioperubatan. Objektif kajian ini adalah untuk mengkaji kesan masa anodisasi dan peningkatan bioaktiviti. TiO_2 nanotiub dicapai dengan anodisasi elektrokimia. Nanotiub dengan saiz diameter dan panjang yang berbeza ditanam pada substrat Ti-Nb dengan pemalar 10 V dan 30 V pada 5 min, 35 min dan 65 min melalui anodisasi elektrokimia. Pengimejan mikroskop elektron (SEM) mengimbas menunjukkan bahawa struktur nanotubular sejajar menegak terbentuk pada permukaan substrat aloi Ti-Nb. Voltan dan masa anodisasi mempengaruhi diameter dan panjang tiub nano. Analisis XRD mengesahkan bahawa terdapat pembentukan lapisan TiO2 dan titanium fasa β. Perbezaan diameter dan panjang tiub nano boleh menjejaskan sifat aloi Ti-Nb seperti rintangan kakisan dan mekanikal. Anodisasi pada 30 V selama 35 min dan 65 min menghasilkan tiub nano dengan diameter 45.53 nm dan 50.66 nm. Anodisasi pada 30 V selama 35 min dan 65 minit menghasilkan tiub nano dengan panjang 1225.77 nm dan 2250 nm. Anodisasi pada 10 V selama 35 min dan 65 min pula menghasilkan tiub nano dengan diameter 20.18 nm dan 21.64 nm. Panjang tiub nano ialah 106.98 nm untuk anodisasi pada 10 V selama 35 min dan 201.13 nm selama 65 min. Permukaan Ti-Nb dianodkan pada 30 V digalakkan meningkatkan pertumbuhan sel apabila masa anodisasi meningkat. Penggunaan tiub nano TiO2 meningkatkan lekatan sel yang penting dalam aplikasi bioperubatan.

CHAPTER 1

INTRODUCTION

1.1 Background study

In recent years, the number of patients requiring artificial alternatives or implants to replace failed tissue, such as arthroplasty, hip joints, craniofacial, maxillofacial, dental implants, prostheses, and surgical instrumental applications, has increased (Goncalves et al., 2020). According to researchers, demand for hip and knee arthroplasties could exceed 3.48 billion surgeries in 2030, up 673 % from 2005 in the United States (Kurtz et al., 2007). Thus, many efforts have focused on identifying acceptable biomaterials for the manufacture of long-lasting medical implants (Khosravi et al., 2020).

Biomaterials are being utilised to replace biological structures due to their potential to improve human longevity and quality of life, which has sparked a lot of interest and quick development. Biomaterial manufacturing, characterization, and application is a fascinating topic that has sparked a lot of research in recent years. Although there are disagreements about what constitutes a biomaterial, they can be defined as implantable medical devices made of metals, ceramics, and synthetic polymers, biopolymers, self-assembled systems, nanoparticles, carbon nanotubes, and quantum dots, drug and gene delivery systems, tissue engineering and cell therapies, organ printing and cell patterning, nanotechnology-based imaging and diagnostic systems, and microbial biomaterials (Williams, 2009).

The restoration of bone tissue, often known as orthopedic biomaterials or implants, is one of the applications of biomaterials (Navarro et al., 2008). These biomaterials must have the appropriate mechanical and biological characteristics.

Metal implants are chosen for load-bearing applicationss; thus, they should have adequate fatigue strength over other biomaterials such as ceramics, polymers, composites, and natural goods. The key requirement for metals to be classified as biomaterials is that they cause no adverse reactions when utilised in the intended biomedical application; in other words, they must be biocompatible.

Mechanical parameters of metal implants include Young's modulus, tensile strength, ductility, fatigue life, and wear resistance (Niinomi, 2008). They also offer outstanding chemical qualities such as corrosion resistance, biocompatibility, and the ability to merge and harmonise with other implant materials (Niinomi, 2008). Because of these characteristics, they are more suited for long-term use in hard tissue applications like hip and knee joints. Metallic biomaterials, such as 316L stainless steels, Co–Cr-based alloys, titanium and its alloys, offer desirable qualities among many types of materials (Sarraf et al., 2014).

Since the 1920s, grade 316L stainless steels (18Cr–14Ni–2.5Mo wt%) have been utilised as implants. The "L" in 316L stainless steel implies a low carbon content, which helps to prevent chromium carbides from forming and improves corrosion resistance. However, stress corrosion cracking, which is unavoidable in 316L stainless steel, might be induced by a combination of tensile stress and a Cl-rich environment, such as the human body fluid, resulting in an unfavourable abrupt implant failure under stress (Xu et al., 2018). Furthermore, while Co–Cr-based alloys have a stronger corrosion resistance in human body fluid than 316L stainless steels, wear and corrosion release some undesirable ions such as Cr and Co (Yamanaka et al., 2019).

Many animal studies have found Co to be carcinogenic, and there have been reports of neurological problems in patients following implantation. Because the released Cr could cause oxidative reactions in blood cells, kidneys, and liver, Co–Cr-

based alloys and 316L stainless steel pose potential dangers as implants. As a result, these two metals may not be the greatest option for orthopedic implants, highlighting the importance of titanium (Ti).

Due to their combination of unique qualities as compared to 316L stainless steel and Co-Cr alloys, titanium (Ti) and its alloys have gained interest for biomedical applications. Ti and its alloys have been used as medical implants due to their long fatigue life, corrosion resistance, high biocompatibility and lower Young's modulus compared to other implants (Su et al., 2018). Titanium and its single-phase alloys α or β and two-phase alloys α + β, which comprise the additions of Al, V, Nb, Ta, Zr, Mo, Si, Sn, Pd, Fe, and Hf and exhibit osseointegrative properties, are the most promising category of biomaterials for such applications.

The long-term implant market has so far been controlled by the two-phase Ti–6Al–4V alloy. Furthermore, there are some clinical concerns about the widely used Ti-6Al-4V alloy. This is because vanadium is a microelement that is present in trace amounts in the human body and is required for normal functioning. A high quantity of V in the body, on the other hand, can cause allergic reactions, renal damage, and digestive and respiratory system discomfort. An excessive amount of V can be harmful to the neurological system and brain cells, resulting in manic-depressive psychosis. Vanadium can also inhibit several enzyme systems and cause genotoxicity, which can harm foetal reproduction and development at various stages. Not only that, after a lengthy period of use, Al has been deemed possibly hazardous and carcinogenic. Furthermore, the Young's modulus of Ti and Ti-6Al-4V is much higher than that of bone, resulting in bone resorption and subsequently, revision surgery (Panigrahi et al., 2016; Okazaki et al., 2005).

Thus, new β-type Ti alloys have been produced that comprise non-toxic elements (Nb, Ta, Mo, Zr, etc.) and have a significantly lower Young's modulus (Niinomi et al., 2012). Ti-Nb based alloys have gotten a lot of interest for loadbearing implants because of their decreased Young's modulus, as well as their superior corrosion resistance and biocompatibility (Niinomi et al., 2012). As a β-stabilizer, niobium has outstanding mechanical, chemical, and biocompatible qualities, allowing it to be used in corrosion-resistant structural materials and biomedical implants.

Electrochemical anodization is a versatile technology that has been used to build thick and homogeneous oxide layers on metals for decades and has demonstrated significant benefits in improving the biocompatibility of metallic implants (Yao & Webster, 2006). When metal is used as the anode in an electrochemical cell, a protective oxide layer with a controlled and desired thickness forms spontaneously. The type of electrolyte, applied current density, electrolyte concentration and temperature, agitation speed, and cathode-to-anode and surface-to-area ratios can all be adjusted to modify the structure of the oxide layer (Ali et al., 2011).

Apart from pure Ti, nanotubes on the surface of binary titanium alloys such as Ti–Zr, Ti–Ta, Ti–Al, Ti–Mo, and Ti–W have piqued interest. The alloy element concentration was connected to the diameter, shape, and distribution of metal oxide nanotubes produced on the surfaces of these alloys. Only a few studies have looked at the production and bioactivity of oxide nanotubes on Ti–Nb alloys. Thus, Ti–Nb alloys containing 40 wt% Nb have recently been identified as new materials for biological applications.

1.2 Problem statement

Currently, 70%-80% of bone implants are made of metallic biomaterials, in particular, titanium and its alloys such as Ti-6Al-4V due to their superior corrosion resistance and biocompatibility, low density, low elastic modulus, and high strength. However, recent studies have demonstrated some limits in biocompatibility due to the presence of toxic Al and V. So, because of that, it can provide accelerated corrosion with consequent damage to the metal implant.

Ti alloys is acknowledged as biocompatible material. But because of the the presence element in the Ti-6Al-4V, another Ti alloys is introduced such as Ti-Nb. However, Ti alloy are nonbioactive and lack rapid tissue integration, which results in the subsequent development of interfacial fibrous tissue, finally leading to isolation of the implants (Das et al, 2008). Therefore, to improve the bioactivity of titanium, it is essential to develop surface modification technologies such as the formation of nanotubes on Ti alloys surface. Titanium alloy surface modification already extensively studied in the view of osseointegrative behaviour.

The electrochemical anodization is growing the TiO_2 nanotubes but the nanotubes can be controlling the diameter and length with voltage and time applied on anodizing. TiO_2 nanotubes was found that it is very efficient in improve bioactivity by formation of hydroxyapatite compared with compact oxide layers. Even though, TiO_2 nanotubes is can improve the bioactivity by the formation of hydroxyapatite, but the diameter and length of nanotubes is affecting the bioactivity. Some researcher had do nanotube on TiO_2 and the length of nanotube is long and bigger diameter which is exceed 3000 nm for the length and 100 nm for the diameter. The hydroxyapatite is difficult to form with the bigger diameter and long of length nanotube.

1.3 Objectives

a) To study the effect of the anodization time and the enhancement of the bioactivity

1.4 Thesis organization

This thesis consist of five chapters as follows:

Chapter 1 consist of introduction of the thesis. It covers background study, problem statement and research objectives.

Chapter 2 consist of literature review. It includes the general review of Ti alloys, fabrication of Ti alloy, preparation of nanostructure on Ti alloy and bioactivity of Ti alloy.

Chapter 3 consist of methodology. It covers the specification of materials and instrument used in this project also the testing conducted.

Chapter 4 includes the result and discussion. It contains the figure, table and chart which represent the experimental result and discussion correspond to the result obtained.

Chapter 5 presnt the conclusion and future scope where a few suggestion for future studies are given for references to other researcher.

CHAPTER 2

LITERATURE REVIEW

2.1 Introduction

Metal implants are the best choice for the long-term replacement of hard tissue, such as hip and knee joints, because of their excellent mechanical properties. Titanium and its alloys are commonly acknowledged as biocompatible metal implants due to their self-organized oxide layer that shields the surface from corrosion and inhibits ion release. They have to be accepted by bone cells, bond with them, and grow on top of them in order to stop the bone from becoming loose. Surface modification is necessary in order to facilitate the osseointegration of these biomaterials into the patient's body.

Anodization, a process in which nanotubes are fabricated on the surface of metal implants, is currently receiving an increasing amount of attention for the purpose of surface modification. This is because electrochemical anodization is a process that produces a durable and corrosion-resistant anodic oxide layer on a metal surface that protects the inner bulk metal (Das aet al., 2008). The nanotube form, in particular, improves the physical interlocking of osteoblast cells on the implant's surface. Additionally, the growth of a bone-like apatite layer promotes osseointegration when nanotubular arrays are formed. After implantation, the diameter and shape of the titania nanotubes have an effect not only on the crystalline size but also on the thickness of the bone-like apatite that forms on the surface of the nanotubes.

2.2 Bone Implant Material

As with other components of our body, bone can become harmed or weakened as a result of aging, trauma, or illness. This damage, which can include bone fractures, low back discomfort, osteoporosis, scoliosis, and other musculoskeletal disorders,

typically happens in elderly persons but is not limited to only those of that age group. Weiner et al have described the basic bone composition that consists mostly of fibrous protein collagen, carbonated apatite (Ca5(PO4, CO3)3(OH)) and water. Over time, both the size of the crystals and the proportions of their components will shift. As a direct consequence of this, bones of younger age replace bones of older age. It shows that bone is tissue that alive and growing.

In the process of biomineralization, bone-forming cells such as osteoblasts are accountable for the generation of the synthesis and deposition of calcium phosphate crystals. These crystals are necessary for conferring hardness and strength on the mineralized structure. Bone was described as a solid material that is extremely porous on a microscopic scale. Because of these pores, bone has a viscoelastic quality. These pores are filled with fluid and cells, including osteoblasts, osteoclasts, osteocytes, and bone lining cells, all of which are capable of regeneration.

The metal implant needs to be bioinert with highly corrosive and demanding environment of the human body. Biomaterials have progressed through three generations from the first, whose particularities were matching its physical properties as a tissue replacement with the least toxicity or biological inertness to the second generation that demonstrate bioactive behaviour (Hench and Polak,2002). Surface treatments have been used to improve the bioactive nature of these biomaterials, especially metals such as titanium and titanium alloys that change the physico-chemical, mechanical and electrical properties of their surfaces (Liu et al,2004). The third-generation biomaterials are intended to promote specific cellular responses at the molecular level (Hench and Polak,2002). Stainless steels, titanium and its alloys, and cobalt chromium alloys make up the majority of the metal implants used today. The only stainless steel that is biocompatible is SUS 316L, which has a Young's modulus

of roughly 160 GPa. However, it has low wear resistance when compared to other metal implants.

Damage bone tissue is usually substituted with an artificial replacement to reinstate its functionality. Biomaterials such as implants are used for repairing injured bones, cartilage or ligaments and tendons (Navarro et al,2008). When producing a bone implant, it is of the utmost importance to choose the appropriate material to use so that the implant itself will be successful. There are several criteria that must be met, including biocompatibility, mechanical properties, high corrosion resistance and osseointegration.

2.3 Biomaterials overview

Biomaterials are synthetic materials that are used in the construction of devices that are intended to replace parts of the human body or for medical appliances. Some examples of these types of devices and appliances include tooth fillings, bone plates, cardiovascular devices, and reconstruction of the maxillofacial region. The use of biomaterials has a number of benefits, some of which are listed below: (i) a lower risk for the transmission of illnesses, (ii) a reduced risk of infection, and (iii) the availability of a large number of materials for a variety of applications that are highly specialized. Studies on biomaterials were first conducted approximately fifty years ago.

Recent developments include bioactive and porous biomaterials with improved capability to facilitate new bony tissue in-growth after implantation (Hench and Polak, 2002). Understanding multiple fields is necessary for the design of biomaterials; these fields include engineering, materials science, and clinical medical sciences. Polymers, metals, ceramics, and composites of these materials are all

examples of specific types of biomaterials. Each of these types of biomaterials is created with a particular functioning in mind.

Long-term clinical performance with no need for revision surgery is the goal that should be aimed for with future implants. It is preferable to avoid having revision surgery due to the negative effects it can have on one's health, social life, and financial situation; also, revision surgery might be more complicated than the first operation. The primary factors that lead to the need for revision surgery include inadequate osseointegration at the interface of the implant and the bone, aseptic loosening, and infections. The aseptic loosening that can happen after long term implantation is primarily caused by I the biomechanical mismatch of the implant and the surrounding tissues, and (ii) the growth of fibrous tissue that causes implant movement. One option for resolving the biomechanical mismatch is to reduce the elastic modulus of the biomaterials by either inventing new materials or integrating porous structures into the biomaterials themselves.

In recent years, a significant amount of research and development effort has been put into the creation of porous biomaterials. Because of its porous form, the artificial implant can more securely hold the patient's bone as it grows into the spaces between the prosthetic. The networked structure will contribute to the promotion of cell adhesion as well as the maintenance of cell growth. Cells are more likely to attach themselves to and proliferate on surfaces with a high surface area. The rough surface of porous materials also provides a system where stress can be transferred from implant to bone (Levine et al., 2007). The studies that were done to fabricate porous biomaterials resulted in several improvements. These advances included not only reaching the required elastic moduli, but also enabling body fluid to flow inside the newly implanted device thanks to the porous structure.

A summary of the most commonly used materials for biomedical application and their advantage and disadvantages is given in Table 2.1. As potential bone transplant scaffolds, porous ceramics such as calcium phosphate and bioactive glass have been investigated, and they have shown outstanding levels of bioactivity, osseoconductivity, and biocompatibility. However, low strength, brittleness, inelasticity, low impact resistance and low toughness of these ceramics are a hindrance for use as bone implants (Chen et al, 2011).

Table 2.1 Materials used in biomedical application (Kun Aussieanita,2014)

Material	Advantage	Disadvantage
Metals: Titanium and alloys Tantalum Magnesium Stainless steel Cobalt chromium	• Biocompatible • Light weight • Cytocompatibility • High corrosion resistance (Ti and Ta) • Biodegradable (Mg) • Excellent mechanical strength suitable for load bearing • Ductile	• No direct bonding to tissue • High elastic modulus • Low corrosion resistance (Mg)
Ceramics: Bioglass Hydroxyapatite Aluminium Oxide	• Biocompatible • Bioactive • Strong in compression	• Brittle • Weak in tension • Low impact resistance
Polymers: Polyester PUL PMMA PLLA PEG	• Biodegradable • Ductile • Easy to fabricate • Light weight	•Low mechanical strength • Bioinert

A favourable environment for cell adhesion, proliferation, and differentiation is provided by porous polymers such as polyurethane (PUR), poly (lactic co-glycolic acid) (PLGA), polylactide (PLLA), and poly-DL-lactide (PDLLA). These polymers are also appealing prospects for usage as implants. Studies of polyesters as biomaterials for bone regeneration applications have increased due to their history of clinical success (Pantojas et al., 2009). Because of their adaptability and ability to control the degradation rate through copolymerization, biodegradable polymers are also advantageous for use in scaffolding. However, their bioactivity as well as their mechanical strength make them unsuitable for load bearing applications.

Metals and their alloys have high mechanical strength, making them appropriate for use as load-bearing bone substitutes. Some examples of such metals and alloys include titanium, stainless steel, and chromium cobalt. In addition, metallic-based biomaterials have a high tensile strength, which is an attribute that polymers and ceramics do not have. Because of this, they are preferable for applications that include load bearing.

2.3.1 Biomaterial: requirement

2.3.1(a) Biocompatibility

Biocompatibility is a required property for materials that come into contact with living tissues and organs. Biocompatibility can be defined as the behaviour of any given material with living tissue and cells including toxic, harmful, or irritative effects (Ibrahim et al, 2017). When a material is implanted in the body, its interaction with body fluids, proteins and cells produces a series of reactions that determines whether or not it is accepted by the host system without causing inflammatory or allergic reactions (Williams, 2008). The main issues regarding biocompatibility are thrombosis and fibrous tissue encapsulation of the implant (Geetha et al, 2009). In order to

determine how the body will respond to a given implant, it is important to understand how the biomaterial surface interacts with it. In addition, to evaluate the overall biocompatibility of an implant, it is important to consider all the elements constituting it, since they could be released into the human body due to wear or corrosion and potentially cause adverse effects (Li et al, 2014).

The human body possesses a remarkable aptitude that allows it to determine whether or not an object is foreign to its system. This is part of the body's protection that against the aggression of the outside organism. When a foreign material is introduced into the body, the immune system will respond. An object is considered biocompatible when it is able to enter the body without causing an immunological reaction. In order for a device to be biocompatible, it must follow a very stringent set of demands from the body (Suvaneeth & Nair, 2018). Biocompatibility term is referring to the biomedical device that interact with the body's tissue which are depends on several factors like the chemical and physical nature of its components, the types of patient tissue that will be exposed to the device and the duration of that exposure.

Material that biocompatible can be referred to as biomaterial because the material is constituted part of medical implant as well as in every aspect of patient health care. The National Institutes of Health Consensus Development Conference of November 1982 defined a biomaterial as any substance other than a drug or combination of substances, synthetic or natural in origin, which can be used for any period of time, as a whole or as a part of a system which treats, augments, or replaces any tissue, organ, or function of the body (Boretos and Eden, 1984). The advantages of using biomaterials are lower risk for transmission of disease, reduced risk of

infection and the availability of many materials for a variety of specialized applications.

2.3.1(b) Mechanical properties

The mechanical properties of the materials are very important in determine the acceptance of foreign materials in body's tissue. When metal implant interacts to the body's tissue, the material must be high in strength and low elastic modulus. The high strength is very necessary in order for the implant to be able to withstand the loads that are place on it. To ensuring the long-term success of the implant, it must have suitable fatigue strength to withstand repeated cyclic loading over time. Low elastic modulus is necessary as bones have a Young's modulus varying from 4 GPa to 30 GPa, depending on the type of bone and the direction of measurement (Murphy et al, 2016). The bone tissue will be unstressed if the implant material is stiffer than bone. This is because the higher stiffer implant will control the majority load and avoid the damage on bone tissue. This condition is called 'stress shielding effect' and leads to bone resorption around the implantation site and consequent disuse atrophy (Chiara Micheletti, 2018). Stress shielding effect is a condition where the bone is loss in the vicinity of implants.

2.3.1(c) Corrosion resistance

Implant materials need to have crucial properties, including resistance to corrosion. Due to the fact that the internal partial pressure of oxygen in the human body is approximately one quarter of the oxygen pressure in the atmosphere, oxidation is less reactive inside the body than it is outside. The human body is a very complex, so the metal implant must be good in corrosion resistance. This is because human body constitutes an aggressive corrosive environment for the implant which is exposed to body fluids that contain different type of corrosive substances. Usually when the

corrosion of implant occurs due to the body fluids, it will discharge unwanted non-biocompatible metal ions. Because of that, the implant's lifetime will shorten or decrease hence the second surgery is needed which is causing the patient is more suffered. Besides that, the occurrence of corrosion resulting in reduce the patient life expectancy and being threatened. The corrosion product or dissolved metal ions can either build up in tissues next to the implant or may be transferred to the other body parts (Okazaki & Gotoh, 2005). Therefore, high corrosion resistance is very important characteristic to avoid the formation of toxic substances which will leads to the development of infection. Furthermore, loss of implant material can result in poor implant host tissue interaction and implant loosening (Geetha et al, 2009).

In terms of their mechanical properties, biomaterials such as stainless steel, cobalt-based alloys, and titanium-based alloys are advantageous. This is because they are able to withstand stresses and undergo plastic deformation before failure. The biocompatibility of metallic implant biomaterials can be determined by the corrosion resistance of the material as well as the biological effects of metal ion release. It is anticipated that metallic implants will be composed of non-toxic elements and will not result in any substantial inflammatory or allergic response in the human body. The implant is stress-shielded and it will deteriorates, this weaken the bone's interface or the implant (Chen and Thouas, 2015). The amount of metal ions that are released by metallic implants should be at their lowest level in the body even under extreme conditions. The interaction of stress shielding, wear debris, and motion at an interface is very destructive and frequently hastens the failure of an implant. There are also worries regarding the elements that are emitted from cobalt-based alloys, as it has been discovered that Ni, Cr, and Co are all harmful substances. They have the potential to

induce systemic allergic reactions in the body of the host, which can lead to an increase in inflammation (Chen and Thouas, 2015).

2.3.1(d) Osseointegration

Bone fusing to titanium was first reported in 1940 by Bothe. Brånemark began extensive experimental studies in 1952 on the microscopic application of bone marrow healing. In early 1960, dental implant application has been introduced because of the Brånemark studies which is established implant integration in dogs without substantial adverse effects to either hard or soft tissue. First defined of osseointegration by Brånemark is vital bones must have some direct contact with the surface of an implant at the light microscopic level of magnification.

In order to meet the fundamental criteria, bone implants need to have the capability of achieving optimum osseointegration. The term osseointegration is to described the condition and the process for having a stable loaded implant in direct contact with bone (Brånemark et al, 1983). To avoid its loosening over time, an implant needs to be able integrate with the bone tissue.

The implant that keeps in bone is to support the artificial process. As example, when lose a tooth, the implant can keep in the bone to support the tooth. When implant is kept in the human bone, there should be interlocking between this implant and the bone because it will be a lot of bearing area so there has to be a bonding between the bone and metal implant. Otherwise, it will not be a good prognosis of the process. Ultimately, it will remove from the bone because of the fibrous interlocking between implant. If there is no proper interlocking, there is no proper integration between bone and implant which will conduct to failure. Good osseointegration is happen when the bone and implant success join together without interposition of a fibrous connective tissue. The factor that effects the osseointegration is the biocompatibility of implant,

the surface condition of implant and the amount of stress on the implant. So, among the metals, titanium shows the best osseointegration compared to the other metal used for the implant.

2.4 Titanium and Titanium Alloy

Titanium and titanium alloys were first introduced in application of aerospace as structure material. The applications of titanium and titanium-based alloys also have been widely explored in the aviation, marine and military manufacturing industries (Polmear, 2006). Later, titanium alloys were used as dentistry implants in 1950. Since then, titanium alloys have been popular in research to be used in metal implants. These proven applications can be attributed to the distinctive properties of titanium and its alloys, such as high strength to density ratio and high corrosion resistance that enable their use as bone substitutes under load bearing conditions. Moreover, titanium as biomaterial is more favourable for load bearing application because it exhibits high tensile strength. Another advantageous in using titanium is its excellent fatigue properties which is not affected when contact with saline body fluids. Titanium is considered non-toxic, even at high doses, to the human body (N. Moritz et al, 2004).

Titanium and titanium alloys can be classified into three categories according to the phases which are alpha phase (α phase), alpha beta phase (α + β phase) and beta phase (β phase). Their crystal structure and properties are summarised in Table 2.1. If the alpha content is higher than the beta in an (α + β) alloy, then it is termed as a near alpha alloy and vice-versa (Oshida. 2007). Titanium is an allotropic element which it can exist in more than one crystallographic. At room temperature, titanium structure is the hexagonal close-packed crystal structure (hcp) also called as alpha phase. Titanium will transform to the body centered cubic (bcc) structure also known as beta

phase when it is solidifies from liquid or when solid titanium is heated to temperature above 883 ^{o}C. These two-crystal structure are the basis for naming the three classes of titanium alloy which are alpha, alpha-beta and beta (Boyer et al. 1994).

Table 2.2 Classification of titanium and titanium-based alloys (Polmear, 2006)

Alloys	Crystalline structure	Properties	Examples of typical materials
α-type	Hexagonal Closed Packed (HCP)	Superior creep resistance High resistance to plastic deformation and lower ductility Favourable for high and low temperature application	Cp – Titanium Ti-5Al-2.5Sn Ti-6Al-2Sn-4Zr-2Mo Ti-8Al-1Mo-1V
(α + β) type		Mixture of α and β phase Composition of 10-50% β phase at room temperature Good mechanical strength	Ti-6Al-4V Ti-6Al-7Nb Ti-5Al-2.5Fe Ti-5Al-3Mo-4Zr
B-type	Body Centered Cubic (BCC)	Low elastic modulus Excellent work hardening and heat treatment capability Good strength and fatigue resistance	Ti-42Nb Ti-30Ta Ti-13Nb-13Zr Ti-35V-15Cr Ti-8Mo-8V-2Fe-3Sn Ti-50Ta-20Zr

Titanium metal can be alloying with other element by selectively stabilized at room temperature to manufacture stable alpha, alpha-beta and beta alloys. There are

three types of alloying element which are alpha stabilizers, beta stabilizers and neutral. Alloying element that used to stabilize the alpha phase are aluminium, tin, oxygen, carbon, nitrogen, gallium, and zirconium while those used to stabilize beta phase are vanadium, molybdenum, tantalum, niobium, iron and chromium. Now, the researcher has been attracted with tantalum, tin, niobium and zirconium for further studies to produce biocompatible metal implant. These elements have been used due to their non-cytotoxicity, good biocompatibility, high corrosion resistance and their complete solid solubility in titanium (Li et al. 2010).

2.4.1 α and near-α Ti alloy

α Ti alloy is consist of α phase with addition α stabilizer while near α Ti alloy is consist of less amount of β stabilizers which is less than 5 wt%. This type of alloy is having a good weldability even though it is not heat treatable. α type of Ti alloy includes a major class of unalloyed titanium or commercially pure titanium, both of which vary in the amount of oxygen and iron that they contain. Pure Ti is known as cp-Ti which is the purity is 99% of pure titanium.

Unfortunately, because of the HCP structure, the mechanical strength of α type Ti alloy is often limited and rather low while the alloy is at room temperature. This limitation cannot be overcome by heat treatment so α Ti alloy is not a good choice in implant application.

2.4.2 (α + β) Ti alloy

(α + β) Ti alloy are the mixture of α and β phase with addition of α and β stabilizer but in this type of alloy, β stabilizer has a higher content around 4 – 6 wt%. Ti alloys of this type are heat-treatable and have good fabricability (Jung et al, 2013). As a consequence of this, mechanical properties can be improved by heat treatment. This type of alloy has high tensile strength and good creep resistance. (α + β) Ti alloy

also good in corrosion resistance and good osseointegration in the human body because it has rapid formation of TiO_2 film at the human bone interface.

2.4.3 β and near β Ti alloy

Most β Ti alloy contain small amounts of α stabilizer which permit second phase strengthening to high level at room to moderate temperature. Even though there is some α stabilizer but still act as β Ti alloy. β Ti alloy is easily cold formidable because the bcc structure is ductile. β-Ti alloys possess lower elastic moduli, higher hardenability, and improvable ductility and toughness as they can be heat treatable (Chang et al, 2019). The major alloying element for β alloy such as Nb are considered to be very biocompatible compare to α alloy.

2.4.4 Ti – Nb alloys

Recently, Ti-Nb alloys is starting to attract attention in the biomedical application. This is because the β stabilizer which is Nb has low Young's Modulus and do not contain toxic element. The in vitro and in vivo biological evaluation of pure metals of potential alloying elements, including Nb, Ta and Zr, indicated that they are less cytotoxic than Ti alloys or even non-toxic (Niinomi, 2003). Among Ti alloys, β alloys exhibit the lowest elastic modulus and wear resistance as well as a high hardenability (Bania, 1994). Nb is a β phase stabilizer that preserve advantageous mechanical characteristic and simultaneously increasing the ratio of the β/α phase. To achieve a complete and stable β phase in Ti-Nb alloy, the content of Nb that should be included is around 35 – 40 wt%.

2.5 Fabrication of titanium alloy

Fabrication of titanium alloy have two method which are conventional method and advanced method. Conventional method that has been used are powder metallurgy

and casting while for advanced method are selective laser melting and electron beam melting.

2.5.1 Powder metallurgy

Over the decades, powder metallurgy (PM) is popular for biomedical properties due to low-cost manufacturing process when producing high performance Ti alloys. PM is undeniably a promising candidate for low-cost-high-performance manufacturing processes (Blaine et al. 2014).

Powder metallurgy is a method of fabrication in which metal powders are produced and further utilized by pressing and sintering to form useful products as shown in Figure 2.1. For many years, powder metallurgy has been widely used especially in the industrial field and many types of materials can use the powder metallurgy process. Metallic alloy powders are produced when the elemental metallic powders and alloying elements go through the mixing or mechanical alloying process.

Figure 2.1 Powder metallurgy process flow

2.5.1(a) Mechanical alloying

Mechanical alloying (MA) is the term for the processing of metal powders in high-energy ball mills. The technique of mechanical alloying was developed in the mid-1960s by John Benjamin to produced nickel-based oxide dispersion strengthened (ODS) superalloys for gas turbine application. During the past 20 years, many researchers attracted with this technique and they put a lot of attention on it. The processing involves cold welding, fracturing, and rewelding of powder particles in a

high-energy ball mill that resulting in a formation of alloy phases. Mechanical alloying described the process when the mixture of powder of different metals or alloys is milled together. Not only that, the purpose of powder of metal milled together is to produce a solid solution either equilibrium or supersaturated, intermetallic or amorphous phase. Homogeneous alloy is obtained when material transfer is involved in the mechanical alloying process.

Different type of high-energy milling equipment is used to produced mechanical alloy. In this experiment, planetary ball mill has been conducting mechanical alloying process. It is called planetary ball mill due to the planet-like movement of its vial. The planetary ball mill is consisting rotating support disk and a special drive mechanism that causes them to rotate around their axes. The vials that rotating around their own axes will produces centrifugal force and that produces by the rotating support disk both act on the vials content, consisting of the material to be ground and the grinding balls. The centrifugal force eternally acts in opposite direction since the vials and supporting disk rotate in opposite direction. This will cause the grinding balls run down in the inside wall of the vial, followed by the material being ground and the grinding ball lifting off and travelling freely through the inner chamber of the vial and colliding with the opposing inside wall as shown in Figure 2.3 (Suryanarayana, 2003). The grinding ball collide with each other will intensify the impact effect considerably.

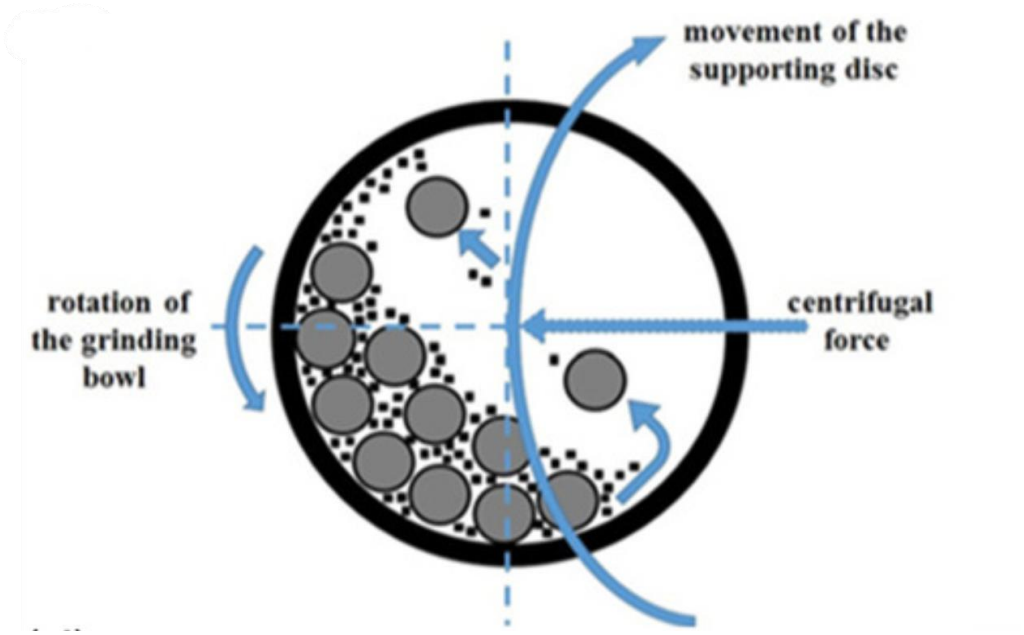

Figure 2.2 Schematic depicting the ball motion inside the planetary ball mill

The grinding balls acquire much higher impact energy than it is possible with simple pure gravity or centrifugal mills. The speed of planetary mill is very important because the impact energy acquired is depend on it and the planetary speed can reach about 20 times the earth's acceleration. The grinding ball will lose the impact energy if the speed is reduced and there is no grinding involve when the energy is sufficiently low. During that time, only mixing occurs in the sample. For titanium, the speed that has been used for grinding is 200 rpm for 2 hours. To get the finer powder, the ball ratio is set to 10:1 which is corresponding to ball and metal powder.

2.5.1(b) Pressing

Powder pressing is the compaction of powders into a geometric form and performed in the room temperature. The purpose of pressing is to obtain required shape and density to make it strong for further process. The pressed powder is known as green compact which has low strength, very fragile and can crumble very easily. The powder must be fed properly into die cavity and suitable pressure is applied to obtain

higher green strength. The density of the green compact depends on the pressure applied. If the pressing pressure is increasing, the compact density approaches that of the metal in its bulk form. Pressing the alloy powder is performed in a rigid die under high pressure which is typically around 135 to 680 Mpa. For titanium, the previous researcher prescribed that the pressure applied is 550 Mpa.

2.5.1(c) Sintering

Sintering is very critical process in the powder metallurgy process because it determines the properties of the alloy. The consolidation of powder compacts into sintered product is accomplished through a process called sintering which make used of heat energy. Sintering is a heat treatment applied to powder compact to impart strength. The temperature that is employed for the sintering process is lower than the point at which the primary component of the powder metallurgy material would melt. The fundamental process that occurs during sintering are densification, grain growth and microstructure evolution. During sintering process, the pores between the powder particle is shrink and cause the grain to arrange into more favorable packing arrangement. Reduced pores and lower interfacial energy between grain boundaries always benefit densification and provide precise control over grain size, sintered density, and phase distribution in the alloy (Kang S J l. 2005). The sintering temperature that suitable for titanium is 1200 ^{o}C for 2 hours (Robert Frykholm and Benjamin Brash, 2015).

2.6 Surface treatment for Titanium alloy

The surface must be modified in order to improve the qualities of an implant and raise the level of its bioactivity when it is in contact with natural tissue. The biological properties of implants can be improved by either modifying the composition