METAL LEACHABILITY, MECHANICAL PROPERTIES AND SURFACE CHARACTERISATION OF Ti6Al4V DENTAL IMPLANTS

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METAL LEACHABILITY, MECHANICAL PROPERTIES AND SURFACE CHARACTERISATION OF Ti6Al4V DENTAL IMPLANTS

by

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LIST OF UNITS AND SYMBOLS

°C	Degree Celsius
g	Gram
L	Liter
μg	Microgram
μm	Micrometer
mg	Milligram
mm	Millimeter
%	Percentage
Rq	Root mean square
Ra	Surface roughness

LIST OF ABBREVIATIONS

Al	Aluminium
AFM	Atomic force microscopy
CAD	Computer-aided design
ICP-MS	Inductively coupled plasma mass spectrometry
SEM	Scanning electron microscope
SBF	Simulated body fluid
Ti	Titanium
Ti-6A1-4V	Titanium, Aluminium, Vanadium alloy
Va	Vanadium

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KELARUTLESAPAN LOGAM, SIFAT-SIFAT MEKANIKAL DAN PENCIRIAN PERMUKAAN IMPLAN PERGIGIAN TI6AL4V

ABSTRAK

Peranti prostetik yang ditanamkan telah menjadi modaliti terpadu dalam pergigian restoratif. Risiko biologi yang berkaitan dengan zarah ion yang dilepaskan dari implan logam adalah kritikal dan sangat dicari. Kajian ini terbahagi kepada *in vitro* dan *in vivo*. Secara spesifik, bahagian *in vitro* adalah untuk menilai kemungkinan kebolehlarutan logam dari implan gigi Ti6Al4V dalam simulasi cecair badan pada pH yang berlainan. Objektif khusus untuk bahagian *in vitro* adalah untuk menganalisis kebolehlarutan logam daripada implan gigi dalam simulasi cecair badan (*SBF*) dalam pH berbeza (3.0, 5.5, 6.5, 7.3,7.5 dan 7.8) menggunakan *inductively coupled plasma mass spectrometry* (ICP-MS), kakisan elektrokimia dan penilaian sifat mekanik. Objektif khusus untuk bahagian *in vivo* adalah untuk menganalisis kebolehlarutan logam dari implan gigi dalam darah, air liur dan sifat mekanik dan perubahan permukaan implan yang gagal. Berdasarkan kajian *in vitro*, terdapat kebolehlarutan logam yang ketara pada akhir bulan pertama hingga ke enam berbanding dengan garis dasar bersama dengan perubahan permukaan yang menyebabkan korosi lubang pada pH yang berlainan.Kawasan hitam di permukaan implan menunjukkan kawasan yang mudah terkena kakisan di bawah analisis SEM. Tidak ada perubahan yang signifikan dalam kekasaran permukaan (Ra) dan kuadrat akar (Rq) berdasarkan analisis AFM. Lapisan oksida permukaan implan Ti6Al4V mengalami tahap kakisan yang berbeza-beza berdasarkan analisis kakisan elektrokimia. Analisis elemen terhingga menunjukkan bahawa peningkatan diameter dan panjang implan penting dalam

taburan tekanan dan diameter memainkan peranan penting dalam taburan tekanan berbanding panjang implan. Keputusan *in vivo* analisis keratan rentas sepuluh subjek implan yang dirawat dengan sepuluh subjek tanpa implan menunjukkan perubahan yang signifikan pada tahap darah titanium dan vanadium pada subjek lelaki dan hanya pada tahap vanadium dalam kumpulan subjek perempuan. Kebolehlarutan titanium dan vanadium dalam air liur adalah signifikan pada kumpulan ujian pada lelaki dan wanita. Secara prospektif, peningkatan yang signifikan untuk kebolehlarutan titanium, aluminium dan vanadium pada tiga dan enam bulan dibandingkan dengan dasar ke atas lima belas subjek lelaki dimana dapat diperhatikan juga dalam kalangan subjek wanita. Titanium, aluminium dan vanadium juga terdapat dalam air liur ketika dasar dan setelah enam bulan dibandingkan. Analisis Soderberg dan Goodman tentang keletihan hidup menggunakan perisian analisis elemen terhingga menunjukkan bahawa implan mempunyai jangka hayat 1×10^6 kitaran dan dapat bertahan hingga satu kali sepuluh. Implan yang dikeluarkan dari pesakit akibat kegagalan implan menunjukkan bahan protin dan biji-bijian yang mendalam menunjukkan kemungkinan kebolehlarutan logam. Terdapat peningkatan yang ketara dalam kekerasan dan penurunan pemuatan mampatan pada implan yang gagal. Dalam batasan kajian semasa, dapat disimpulkan bahawa kebolehlarutan logam berlaku daripada implan gigi.

METAL LEACHABILITY, MECHANICAL PROPERTIES AND SURFACE CHARACTERISATION OF TI6AL4V DENTAL IMPLANTS

ABSTRACT

Implanted prosthetic devices have become an integral modality in restorative dentistry. Biological risk associated with ionic particles released from metallic implants are critical and very much sought after. This study is divided into *in vitro* and *in vivo*. The main objective of the study was to evaluate metal leaching from dental implants Ti6Al4V. Specifically, the *in vitro* part aimed to analyse metal leaching from dental implants in simulated body fluid (SBF) at different pH (3.0, 5.5, 6.5, 7.3,7.5 and 7.8) using inductively coupled plasma mass spectrometry (ICP-MS) with electrochemical corrosion and mechanical properties evaluation. The specific objective for the *in vivo* part were to analyse metal leaching from dental implants in blood and saliva, mechanical and surface changes properties of failed implants. Based on *in vitro* study, significant metal leaching occurred at the end of first until six months compared to baseline along with surface changes leading to pitting corrosion at different pH. Black areas on implant surfaces suggesting areas prone for getting pitting corrosion under SEM analysis. There were no significant changes in surface roughness (Ra) and root mean square (Rq) based on AFM analysis. The surface oxide layer of Ti6Al4V implants can undergo varying degree of pitting corrosion as shown by electrochemical corrosion (ECC) analysis. The finite element analysis (FEA) showed that increasing both diameter and length of implant are important in stress distribution and diameter play a salient role in stress distribution compared to implant length. In vivo cross-sectional analysis of ten implant treated subjects with ten without implants showed significant change in titanium and vanadium level in blood among male subjects and only in vanadium level in female subjects. Titanium and vanadium level in saliva was significant in test group for both male and female. Prospectively, a significant increase of titanium, aluminium and vanadium level in blood at three and six months compared to baseline among male subjects which was also observed among female subjects. Titanium, aluminium and vanadium were also expressed in saliva at baseline and after six months. Soderberg and Goodman analysis of fatigue life showed that the implant has a life of 1×10^6 cycle. Implants removed from patients due to implant failure revealed proteinaceous material and deepened grains suggesting chances of metal dissolution by SEM analysis. Failed implants showed significant increase in hardness and decrease in compression loading. Within the limitation of the current study, it can be concluded that metal leaching does take place from dental implants.

CHAPTER 1

INTRODUCTION

1.1 Background of Study

Implants were used in ancient China as early as 2000 before century (BC) (Lasemi & Navi, 2020). Implanted prosthetic devices have become an indigenous treatment modality in fixed restorative dentistry. Dental implants are utilised to restore missing natural teeth in partial and completely edentulous situation. Improvement in health care and increased life expectancy of population demands implants with minimal deleterious effects (Garçon et al., 2016). Finding by American Academy of Implant Dentistry (AAID) highlighted that three million people in the United States have dental implants, and new cases of 500,000 implants placement are recorded each year (Shavit et al., 2019). Longevity and complications remain as significant issues and provide opportunities for the creation of improved devices.

A dental implant is a good illustration of the integrated system of science and technology for replacement of missing natural tooth (Oshida et al., 2010). A dental implant is an invasive component that is placed within the bone of the jaw to support a dental prosthesis such as a crown, bridge, denture, facial prosthesis and orthodontic anchor (Warreth et al., 2017, Oshida et al., 2010). Completely edentulous patients should be made aware that the resorption will continue with significant risk for removable denture, resulting in the need for bone grafting procedures depending on the individual clinical scenario for placement of implants later. Although fixed partial dentures and dentures address the cosmetic problems of missing teeth, the choice depends on the cost and knowledge of treatment (Quran et al., 2011). Fixed partial denture treatment involves the reduction of natural teeth for the placement of crown either in metal or ceramic (Tinschert et al., 2001). Dental implants on the other hand exert appropriate force on the jaw bone and keep them functional and healthy.

Implant supported prosthesis are reported to improve phonetics, masticatory movements and aesthetics (Warreth et al., 2017). Implant offer a predictable treatment course as it allows teeth replacement without coronal tooth preparation (Levin, 2008). A tremendous growth in dental implant market has been observed in last few years (Rao & Bhat, 2015). Increased use of implants results from many reasons which include the need for implant treatment among edentulous subjects associated with aging and increased life expectancy (Emami et al., 2013). A non-smoking individual of normal weight is having life span of 85 years (Misch, 1999) which emphasises the need for long term evaluation and effects of implants in the body.

Success of a biomaterial is defined in terms of years of reliable service rather than a lifetime of device functionality (Park & Lakes, 2007). Corrosion of dental restoration is a pertinent issue as it is intended to function in human body for a life time. It is a progressive deterioration of metals by electrochemical attack when it is subjected to the electrolytic domains provided by the human body (Olmed et al., 2009). Corrosion is of grave concern as metal ions and debris are produced in this process and the aggregation of this may cause detrimental tissue reactions *in vivo* (Eltit et al., 2019).

1.2 Research Problem

The province of biomaterials has become an indispensable area, as they can augment the standard and endurance of life expectancy. The science and technology related to this has now give rise to multi-million dollar business (Manivasagam et al., 2009). Biodiversity demands biomaterial with improved elemental properties such as mechanical and biological properties which includes modulus of elasticity closer to bone, improved corrosion resistance and osseointegration (Al-Zubaidi et al., 2020). The oxide layer formed on the implant surface can range from 2-7 nm thickness (Oshida et al., 2010). It plays a decisive role in biocompatibility as the presence of an oxide film causes less dissolution rate of passive metal at a given potential (Rakic et al., 2016). Implants when exposed to blood and body fluid will be able to induce implants corrosion due to the presence of amino acids, and ions like calcium, potassium, magnesium and zinc. There will be imbalanced of equilibrium between electrons on the implant and cations in the solution as the constituents of blood tend to secure the metal ion and carry them away (Eliaz, 2019). Those implant areas adhered with protein undergoes oxygen deprivation of certain region of implant surface and trigger preferential corrosion of oxygen-deficient regions and breakdown of the passive layer. Either serum or albumin are used for studying the corrosion effects of proteins on implants. Saliva contains enzymes, mucus and bacterial cells of 500 million per mL (Vila et al., 2019) with pH value ranging from 5.5 to 7.5, but under plaque deposits it can be as low as 3 (Vila et al., 2019). Hostile condition of varying pH and stress from masticatory force ranging from 150 to 250 N (Oshida et al., 2010) can affect biocompatibility of metallic materials. Subjects with implants should be curious about the side effects of implant treatment as the total number of implants can be up to eighteen especially in case of full mouth rehabilitation. Hence, there is a need to determine the metal leaching that might be occurring in the dental implants.

1.3 Research Objectives

1.3.1 General objective

i. The study aims to determine metal leaching from detal implants in simulated body fluid, blood and saliva.

1.3.2 Specific objectives

In vitro evaluation of metal leachability, surface changes for corrosion, and mechanical properties included following:

i. To determine metal leaching from dental implants in simulated body fluid (SBF) at different pH using inductively coupled plasma mass spectrometry (ICP-MS) and analysis of implant surface changes using scanning electron microscope (SEM) and atomic force microscopy (AFM).

ii. To analyse the electrochemical corrosion of dental implants in SBF at different pH and implant surface changes using SEM.

iii. To determine mechanical properties in terms of deformation, stresses, strain, fatigue, and surface area changes of dental implants using finite element analysis (FEA).

In vivo evaluation of metal leachability, surface changes for corrosion, and mechanical properties included following:

iv. To determine the level of metal leaching in blood and saliva among dental implant subjects using inductively coupled plasma mass spectrometry (ICP-MS) analysis.

v. To compare mechanical properties in terms of fatigue, microhardness, compression load, and surface changes of retrieved dental implant following implant failure using SEM.

1.4 Research Hypothesis

Null hypothesis:

H_o: Null hypothesis state that there is no metal leaching from implants in SBF, blood and saliva

Alternative hypothesis:

H_a: Alternative hypothesis state that there can be metal leaching from implants in SBF, blood and saliva

1.5 Justification of the study

The cardinal prerequisite of a biomaterial is that the material and the tissue domain of the physique should coexist without having any unwanted effect on each other. Ti6Al4V implant was selected to assess the possibility of metal leaching in this study, owing to the fact that it perpetual to be a key material in orthopaedic and dental applications where high strength is necessary. In 2010, a faulty hip implant was recalled globally due to metal leaching by its manufacturer due to increased failure rate leaving implanted patients with lifetime health sufferings (Ratna et al., 2018). This incident put forward the question on the credibility of implants. There is a need for evaluation of corrosion status of implants owing to the fact that there are more than 500 types of implants available in the market. This project is conducted to explore the possibility of corrosion of a commercially available dental implant in biological environment and finding out methods to identify metal leaching in subjects undergoing implant placement. Appropriate tool to monitor biological risk associated with ions or particles released from metallic implants are yet to be emphasised. This study attempt to evaluate the physical, chemical and mechanical properties of commercially available dental implants using suitable methods.

1.6 Thesis Outline

This thesis entitled metal leachability, mechanical properties and surface characterisation of of Ti6Al4V consists of nine chapters. Chapter 1 is introduction, which explains objectives of the research along with justification of the study. Chapter 2, the literature review narrates the overview of literature related to research objectives. Chapter 3 explains the materials and methods in relation to metal leachability, mechanical properties and surface characterisation. Surface topography of Adin Toureg^{TM} (S) dental implants were assessed using SEM and AFM analyses. Mechanical characterisation assessments involve the finite element analysis of deformation, stress, strain, fatigue life and physical testing of implants. *In vitro* metal leachability involved the electrochemical corrosion evaluation and inductively coupled plasma mass spectroscopy (ICP-MS) analysis of simulated body fluid immersed with implants for six months. *In vivo* analysis included metal leaching analysis and assessment of dental implants following implantation failure. Metal leaching analysis of blood and saliva was performed using ICP-MS. Assessment of dental implants following implantation failure performed using SEM, AFM, Vickers microhardness assessment and compression loading assessment.

Results obtained for six objectives are explained in the following chapters from Chapter 4 to 8. Chapter 4 covers the results for metal leaching from dental implants in simulated body fluid at different pH, the first objective. Chapter 5 explains the results of electrochemical corrosion on dental implants at different pH and surface changes. Chapter 6 covers the results from finite element analysis of dental implants.

Metal leaching from dental implants into blood and saliva results are discussed in chapter 7. Chapter 8 presented the results obtained for assessment of retrieved implants following implantation failure. Chapter 9 covers the limitations of the current study, suggestion for future improvements and conclusions in related work towards Ti6Al4V dental implants.

CHAPTER 2

LITERATURE REVIEW

2.1 Biomaterial used as medical implants

Appropriate selection of biomaterial is a key factor for success of implants. The required properties of biomaterial as implant include osseointegration, biocompatibility, modulus of elasticity closer to bone (18 GPa), improved tensile, compressive, shear strength, yield strength, hardness and fatigue strength to prevent fracture along with 8% ductility (Saini et al., 2015). Stainless steel and titanium were commonly used implant material and zirconia being the recent material of choice (Yeung et al., 2007). The 316 L also known as ASTM138 was the first form of stainless steel used. Metallic implants are prone to undergo crevice, pitting corrosion which drive researchers to look for other replacement material (Eliaz, 2019). Commonly used orthopaedic implant materials are listed below in Table 2.1 (Manivasagam et al., 2010).

Table 2.1 Commonly used orthopaedic implant alloys (Manivasagam et al., 2010)

Alloy designation	**Elastic modulus GPa**	**Yield strength MPa**	**Ultimate tensile strength MPa**
Stainless steel	200	170-750	495-950
Co-Cr-Mo	200–230	275–1585	600–1795
Commercially pure Ti	102	692	785
Ti-6Al-4V	110	850–900	960–970

2.2 Biomaterial used as dental implants

Dental rehabilitation using implants has increased tremendously in last 30 years. Endosteal implants are placed into the bone whereas subperiosteal implants are placed under periosteum when there is no adequate bone to support the implant. Transosteal implants extend from one cortical bone to the other. The electrochemical "inertness" ranking of the metal surfaces tested was increasing in the order of gold, stainless steel, the cobalt-based alloy, and the TiAlV alloy, with the pure metals Ti, Nb, and Ta being the most favourable (Gibon et al., 2017). Basic classification of dental implants is showed in Table 2.2.

Table 2.2 Classification of dental implants (Steigenga et al., 2003)

Design	Attachment	Macroscopy body design	Surface	Material
Endosteal	Osseointegration	Cylinder	Smooth	Metallic
Suberiosteal	Fibrointegration	Thread	Machined	Ceramic
Transosteal		Plateau	Textured	Polymer
Intramucosal		Perforated	Coated	carbon
		Solid		
		Hollow		

Implants made from non-metal materials is usually placed in the aesthetic zone. It is a metal oxide identified in 1789 by German Chemist Martin Heinrich Klaproth. Zirconium oxide implants have outstanding mechanical properties, good stability, a high biocompatibility and a high resistance to scratching and corrosion (Andreiotelli et al., 2009). Implants made of surface treated zirconia possessed high torque removal even though the fabrication of surface modified zirconia implants are difficult. Newer method to improve surface modification of zirconia on the dental implants include CO_2 laser modification (Andreiotelli et al., 2009). However, long term controlled clinical trials on surface modified zirconia of dental implants are still lacking (Apratim et al., 2015).

2.3 Dental implants

Implants have become reliable treatment option for missing teeth after Branemark® has been introduced in the market in 1960. The use of titanium (Ti) biomaterials has revolutionised clinical and oral implantology in which titanium is currently the implant material of choice (Osman & Swain, 2015).

There are many different types of titanium obtainable as implant biomaterials according to American Society for Testing and Materials (ASTM) as displayed in Table 2.3 (Davis, 2003).

Table 2.3 Implant materials commonly used (ASTM International, 2014)

Alloy Designation		**Microstructure**
Commercially pure Ti	Cp Ti grade I	A
	Cp Ti grade II	A
	Cp Ti grade III	A
	Cp Ti grade IV	A
Ti-6Al-4V		α/β
Ti-6Al-4V ELI		α/β
Ti-6Al-7Nb		α/β
Ti-5Al-2.5Fe		α/β
Ti-12Mo-6Zr-2Fe		Metastable β
Ti-15Mo-5Zr-3Al		Metastable β Aged β+α
Ti-15Mo-2.8Nb-3Al		Aged β+α Metastable β Aged β+α

Commercially pure titanium Grade IV contains approximately 0.4% oxygen concentration while Grade I contains only 0.18% oxygen (Saini et al., 2015). Dental implants are made from Grade I to IV which contain 99% pure titanium with 1% other elements like oxygen, carbon, hydrogen, iron, nitrogen. Grade V titanium alloy contains 90% titanium, 6% aluminium and 4% vanadium (Elias et al., 2008). Aluminium and vanadium act as phase stabilisers in Ti6-Al-4V as shown in Table 2.4.

Table 2.4 Alpha and Beta phases of Ti6Al4V (Elias et al., 2008)

Alloy	**Phase stabiliser**	**Phase**	**Formation temperature**	
Ti-6Al-4V	Aluminium	Alpha	Below 882 °C	Hexagonal close packed crystal lattice
	Vanadium	Beta	Above 882 °C	Body centred cubic

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2.3.1 Methods for implant fabrication

Metal implants are typically manufactured by electron beam or selective laser melting method. Additive manufacturing technology allowed fabrication of implants with rapid prototyping or three-dimensional (3D) printing (Sing et al., 2016). Currently standard methodology is yet to be developed for the additive technology as the advantages and disadvantages of this procedure is yet to be defined (Barazanchi et al., 2017).

2.3.2 Designs of dental implant

Implant stability and success are strikingly governed by design of implants. The macroscopic geometric pattern of a dental implant can assume a cylindrical, conical or tapered form combining the advantages of both designs. Advantage of tapered implant primary stability by gradually allowing thin ridge expansion and determining the least stress possible at the interface with the surrounding bone (Moon et al., 2010). Diverging neck geometric walls type seeming to be the best form, as it can provide a slightly higher primary stability after the implant insertion (Tetè et al., 2012). Besides, bone behaviour is same for converging, diverging and straight wall before and after loading of implant (Tetè et al., 2012) Characterisation of implant design are shown in Table 2.5.

Table 2.5 Designs of dental implant (Misch, 2004)

Design	Characterisation	
Macrostructure	Shape of the body	Cylindrical, Conical Tapered Solid, Hollow, vented, Threaded, Plateau, Perforated
	Characteristics of the neck	Straight walls Diverging walls Converging walls
	Number of the thread leads	Single Double Triple
	Shape of the thread	V-shape Square Buttress Reverse Buttress Spiral
Microstructure	Surface treatment	Smooth, Machined, Coated and Textured

A crest module of 20 degree will impose compressive force and decrease the bone loss compared to shear stress of parallel sided crest module (Misch, 1999). This is due to the fact that parallel sided crest module can cause shear stress in the crestal region whereas when crest module is more than 20º increases bone implant contact and produce compressive force . Rough implant surfaces provide more bone implant contact and improved osseointegration compared to smooth implants (Wennerberg et al., 1996). Distance between adjacent thread measured parallel to the axis is called thread pitch. Surface area of threaded implant is around 30% to 500% more than cylinder implants (Misch, 2004).

The outer diameter of thread is usually around 3.75 mm. The threads with a height exceeding 0.44 mm is able to provide excellent biomechanical response when inserted into bone tissue of low density with immediate loading (Misch, 2004). The thread height is defined as the distance between the major and the minor diameter of the thread. A shallow thread depth favours insertion. Deeper threads produce an increase of the surface and indicated in areas of low density bone and high occlusal stress (Tetè et al., 2012). The thread depth of most V-shape thread is 0.37 mm (Misch, 1999). Implant diameters up to 6 mm are available, not used widely because of insufficient bone width in most of the clinical situations due to bone resorption after extraction of tooth (Yamada et al., 2008).

2.4 Considerations for titanium based dental implant placement in human

2.4.1 Diagnostic considerations for titanium based dental implant placement

Diagnosis and treatment planning in implantology involve following steps. Initial steps includes listening to patient wishes and desires along with dental and medical history evaluation (Gowd et al., 2017). Intra-oral examination involves periodontal examination, both gingival and hard bone tissues expressed in Table 2.6. When tissue type is thick corresponding gingiva will be flat with large amount of attached gingiva and thick bone will be present which can resist trauma infrabony pocket. Root dehiscence is common in thin gingiva with thin peridontium.

Table 2.6 Tissue biotype (Becker et al., 1997)

Periodontal tissue biotype	**Thick tissue biotype**	**Thin tissue biotype**
Gingiva architecture	Flat and dense fibrotic	Pronounced scalloped
	Large amount attached gingiva	Minimal amount of attached gingiva
Bone	Thick osseous form resistant to trauma	Root dehiscence and fenestrations
	React to disease with pocket formation and infrabony defect	React to disease with gingival recession

Occlusal analysis was to establish implant protected occlusion. The longevity of the prosthesis planned is influenced by occlusal pattern (Miyata et al., 2000). Restoration size is influenced by drifting of adjacent teeth into the missing area. Diagnostic imaging is performed to evaluate the deficiency of bone and ridge collapse, proximity to nerve bundle or anatomic structures, available bone quantity and quality (Juodzbalys & Kubilius, 2013). A radiographic template is fabricated to determine surgical plan based on diagnostic wax up and proposed future restoration (Lal et al., 2006). Radiographic template is worn at the time of tomography and panoramic imaging. Patient education on the surgical protocol and informed consent is necessary before proceeding with implant surgery (Moy et al., 2005).

2.4.2 Pre-prosthetic considerations in dental implant placement

Surgical placement of implant in this study is done following original Branemark protocol in two stages as it provides an extremely predictable surgical replacement of missing tooth. It involves vestibular incision with two stage surgery. After implant is buried for a period of six months for maxilla and three months for mandible crestal incision is performed for trans-epithelial attachment (Handelsman, 2006). Prosthetic phase is done usually after a healing period of three months, which gives a time period for correction of hard and soft tissue for aesthetics with crestal incision to expose fixture (Engquist et al., 2002). Countersinking is preferred for aesthetic emergence profile and to avoid transmucosal forces by removable appliance (Rosenbach, 2016). The countersink provides adequate space between platform and the definitive prosthesis to have a steady shift and emergence profile (Al-Sabbagh, 2006).

Drilling with adequate irrigation is necessary to avoid overheating of bone in dense bone. Bone damage is reported at temperature above 60 °C (Trisi et al., 2011). Under preparation of implant site bone is preferred in Type 3 and Type 4 bone. Osteotomy technique, by compressing soft bone can attribute to implant stabilisation by using final drill considerably less than implant size to be used. All drilling procedures should be performed at low speed 800 rpm to 2000 rpm, in-out motion. Implant placement should be accomplished at 25 to 30 rpm (Goswami et al., 2015). This is to achieve primary stability by achieving compression around the implant. Higher torque means higher stability. It is the amount of torque required to advance the implant (Goswami et al., 2015). Implant insertion is done by using torque wrench with a torque of 35 Ncm for maximum tightening to prevent screw loosening (Neugebauer et al., 2009). Torque greater 50 Ncm can cause bone necrosis or fracture, and damage of implant (Trisi et al., 2011).

In response to implant biomaterial placement in the human body, foreign body reaction of neutrophils, monocyte derived macrophages, lymphocyte occurs (Anderson et al., 2008). Platelets play a role in formation of growth factors like platelet derived, and biochemical changes such as calcium enhancement, phospholipid hydrolysis, induction of phosphotyrosine are reported (Mavrogenis et al., 2009). At day one following implantation osteoblasts and mesenchymal cells had migrated to the implant surface and forms bone proteins and non-collagenous matrix which influences cell adhesion and mineral binding. Early formed poorly mineralised osteoid later then converted to 0.5 mm thick layer composed of sialoprotein, osteopontin, calcium, phosphorus.

DNA microarray *in vivo* bone healing around titanium implant expressed 86 up regulated genes compared to osteotomy healing group (Kojima et al., 2008). BMP-2 (bone morphogenetic protein 2) are found to enhance new bone growth (Lan et al., 2006). Healing of bone is influenced by 3 genes such as apolipoprotein E, prolyl 4hydroxylase, alpha subunits (Ogawa & Nishimura, 2006). Vertical bone loss in the radiograph should be less than 0.2 mm per annum for a successful implant placement and review (Ramanauskaite & Juodzbalys, 2016).

2.4.3 Prosthetic considerations in dental implant placement

Five prosthetic options were introduced by Misch FP-1, FP-2, FP-3, RP-4, RP-5 as shown in Table 2.7 (Misch 1999).

Table 2.7 Prosthodontics options of implant restorations (Misch 1999)

Type	**Definitions**
FP-1	Replaces only the crown and appears like a natural tooth
FP-2	Replaces the crown and a portion of the root, but elongated or hyper contoured in the gingival half
FP-3	Replaces missing crowns and gingival colour and edentulous site
RP-4	Overdenture completely supported by implants
RP-5	Overdenture supported by both soft tissue and implant

FP=Fixed prosthesis, RP=Removable prosthesis

Biomechanical load on implant is reduced by directing occlusal load axially and reducing cantilever length (Misch, 2004). Small occlusal table should be designed and RP-4and RP-5 is preferred in nocturnal parafunctional cases (Misch, 2004).

2.5 Biocompatibility & cytotoxicity of implant biomaterial

Titanium and titanium alloys are biocompatible owing to the presence of titanium oxide layer (Cui et al., 2005). Vanadium present in titanium alloy is found to be cytotoxic (BomBač et al., 2007). Viability of human gingival fibroblast and human osteogenic cells were reported to be influenced by titanium and zirconium when metabolic activity of osteosarcoma-derived osteoblasts (SaOs-2) and human gingival fibroblasts (HGF) and the cytokine expression of monocytes (THP-1) assessed using the mitochondrial activity and enzyme-linked immunosorbent assays (Schwarz, 2019). It is reported that nanosized (20-250 nm) titanium particles are associated with increased uptake and more cytotoxic effects compared to micro sized (0.3-43 µm) titanium particles (He et al., 2015). He and co-worker (2015) performed cytotoxicity for Ti microparticles (Ti-MPs, <44 µm), NiTi microparticles (NiTi-MPs, <44 µm), and Ti nanoparticles (Ti-NPs, <100 nm) in periodontal ligament (PDL)-hTERT cells measured with XTT test. It has been shown that titanium dioxide particle of 100 nm can be more toxic (Cai et al., 2011) and particles with sizes of 1.5-5.0 µm were clearly seen in cytoplasm (Choi et al., 2005).

Titanium and titanium alloy implants are processed via additive layer and metal injection moulding. Alternative to machining and casting of manufacturing dental implants, newer methods such as additive layer manufacturing or 3D printing and metal injection moulding proved biocompatible along with other advantages like improved designing, declined waste and efforts in manufacturing (Sidambe, 2014). Metal injection moulding technique has advantage of reduced cost compared to Kroll extraction process of titanium raw material. It helps to produce porous implants with better mechanical property such as modulus elasticity close to bone (Sidambe, 2014).

2.6 Effect of pH on the implant surface

Implants in dextrose-containing solutions were more prone to corrosion than those in Ringer's solutions alone. Increasing the acidity also yielded greater corrosion rates for the dextrose-containing solutions and the solutions without dextrose (Tamam & Turkyilmaz, 2014b). The electrochemical corrosion properties of titanium implants were studied in four different solutions: Ringer's physiological solution at pH = 7.0 and pH = 5.5 and Ringer's physiological solution containing 15 mM dextrose keptat above pH. Corrosion behaviors of dental implants were determined by cyclic polarization test and electrochemical impedance spectroscopy. Implant failure is more likely to occur in persons with medically compromising systemic conditions, such as diabetes related to high blood glucose levels and inflammatory diseases related to pH levels lower than those in healthy people.

Coating an implant with a pH buffering agent can induce the attachment of platelets, proteins, and cells to the implant surface when studied as followed. Titanium discs and implants with conventional SLA surface (SA), SLA surface in an aqueous calcium chloride solution (CA), and SLA surface with a pH buffering agent (SOI) were prepared (Pae et al., 2019).

2.7 Titanium based dental implant

2.7.1 Mechanical properties of dental implants

Resistance to corrosion plays an important role in stability and mechanical strength (Darband et al., 2020). Biomaterials with similar modulus of elasticity with bone are recommended since they assure uniform tensile distribution and avoid stress shielding following implant fixation (Ryan et al., 2006). The ductility minimum value is 8%; hardness and tenacity can also be evaluated as ways of assessing the biomaterial response; the increase in hardness reduces the wear incidence (dos Santos et al., 2017). The standard mechanical properties (ASTM, 2014) of commonly used implants are listed in Table 2.8.

Table 2.8 Mechanical properties of implant material (ASTM, 2014)

Alloy		**Microstructure**	**Elastic modulus Gpa**	**Yield strength MPa**	**Tensile strength MPa**
Commercially pure Ti	Cp Ti grade I	α	102	170	240
	Cp Ti grade II	α	102	275	345
	Cp Ti grade III	α	102	380	450
	Cp Ti grade IV	α	104	483	550
Ti-6Al-4V		α/β	110	850-900	960-970
Ti-6Al-4V ELI		α/β	113	795	860
Ti-6Al-7Nb		α/β	105	921	1024
Ti-5Al-2.5Fe		α/β	110	914	1033
Ti-12Mo-6Zr-2Fe		Metastable β	74-85	1000-1060	1060-1100
Ti-15Mo-5Zr-3Al		Metastable β	75	870-968	882-975
		Aged β+α	88-113	1087-1284	1099-1312

The rigidity of implant should be similar to bone for effective distribution of masticatory stress (Shayesteh Moghaddam et al., 2016).

2.7.2 Finite element analysis of dental implants properties

Finite element analysis (FEA) is a numerical analysis of stress to assess problems in bioengineering field to avoid problems before it happens (Yang, 2019). In the first step implant is subdivided into suitable elements of finite dimensions with specific geometric shape of triangle, square, and tetrahedron with a specific internal strain function. Utilising these strain functions, the equilibrium equations between the external force acting on the element and the displacement happening at each node is determined (Geng et al., 2001). It is been postulated that implant length does not decrease the stress distribution of either the implant or the bone. Alternatively when implant diameter increases, this reduces the stresses (Huang & Tsai, 2003). FEA analyses showed that highest stresses were located at loading areas of abutments and cortical bone for all models (Papavasiliou et al., 1996). Electromyographic and occlusal transducer studies have shown that masticatory forces can be 89–150 N at the incisors (anterior region), 133–334 N at the canines, 220–445 N at the premolars (intermediary region), and 400–600 N at the molars (posterior region) (Souza et al., 2015). It is been pointed that implant diameter was reported to be more important than implant length in distributing stresses to the bone in the case of two stage implant treatment (Lee et al., 2005).

It is been suggested that compact bone overloading may occur in compression (due to the lateral components of occlusal load), and overloading at the interface between cortical and trabecular bone can occur in tension (due to the vertical intrusive loading components) (Baggi et al., 2008). It has been pointed that under 100 N occlusal load, implant with 3.5 mm width expressed more stress on implants and abutment when compared to 2.9 mm width for all on four prostheses implants (de Melo Jr & Francischone, 2020).

The stress distribution pattern around dental implant is different compared to natural teeth owing to the absence of the periodontal ligament around dental implants (Baggi et al., 2008). One study pointed out that it is better to reduce the bone height and insert shorter implants with a greater diameter than longer implants with a smaller diameter in terms of stress distribution (Jafarian et al., 2019). Most undesirable situation in both bone and implant are the oblique occlusal loading into a narrow diameter implant when a load of 200 N was applied (Qian et al., 2009).

It is been suggested that the simplest way to improve fatigue resistance is by increasing implant diameters (Duan et al., 2018). Implants with 4 mm diameter have 30% more fatigue resistance than implants with 3.75 mm diameter (Rangert & Forsmalm, 1994). Reduced mechanical properties and fracture of implant can occur due to regular use of mouth wash containing 1500 ppm sodium fluoride solution for five year as a part of oral hygiene leading to implant failure by fracture (Roselino Ribeiro et al., 2007). Implant body/fixture fracture, abutment screw fracture, abutment fracture, fractured prosthesis are the results from mechanical implant failure. There had been 2% incidence of fracture of implants and most of the implants served three to four years before fracture (Goiato et al., 2019).

2.7.3 Physical properties of dental implants

Blood coagulation and platelet adhesion remain major impediments to the use of biomaterials in implantable medical devices. Adherence of the cells to the implant is influenced by physical properties of biomaterial (Table 2.9) (Xu et al., 2014).

Table 2.9 Properties of biomaterial (Xu et al., 2014)

Properties of biomaterial	**Biological response**
Wettability	Blood coagulation
Roughness	Protein adsorption
Chemical composition	Platelet adhesion

Ultraviolet irradiation can cause enhanced wettability thereby hydrophilic nature and protein adhesion (Rupp et al., 2018). It also can cause a reduction in microbial adhesion with a favourable alteration of surface (Desrousseaux et al., 2013). An implant surface with a pit of 1.5 mm in depth and 4 mm diameter is considered as ideal (Le Guehennec et al., 2007).

Corrosion in dental implants are relevant due to dissolution of alloy components and bone destruction due to current flow from galvanic corrosion (He et al., 2008). In most of the cases, cells prefer rough surface to smooth ones, because rough surfaces favour proliferation and also provides increased surface area (Kunzler et al., 2007). The surface roughness of a biomaterial should be studied in terms of amplitude and organisation. Studies revealed that low-amplitude surface grooves induce orientation of groups of cells or individual cells along its axis; this being the basis to state that the best parameter for the orientation of cells would be the groove's width and not its depth (dos Santos et al., 2017). Still, oriented osteoblasts tend to mineralise more quickly, favouring the tissue/implant osseointegration process (dos

Santos et al., 2017). In contact osteogenesis, bone forms on the implant surface, while in distance osteogenesis, the bone grows from the old bone surface toward the implant surface in an appositional manner. Distance osteogenesis is more common in smooth surface, while both contact and distance bone formation are seen in rough surface (Schwartz et al., 2001). However high roughness surfaces could show risks of increased peri-implantitis, peri-implant mucositis, or ionic release from dental implants containing alloys (Delgado-Ruiz & Romanos, 2018).

According to Wennerberg and Albrektsson the optimal surface roughness (golden point) was around 1.5 μm; lower surface roughness give a weaker bone response due to less cell adhesion (Wennerberg & Albrektsson, 1996a). Laser polishing or electropolishing helps produce smooth surfaces with an average surface roughness of 0.5 to 0.8 μm (Table 2.10) (Wennerberg & Albrektsson, 1996b).

Table 2.10 Different scales of roughness of implant surfaces

Types	**Roughness**	**Response**
Macro-roughness	≥ 10 μm	Increased risks of peri-implantitis
Micro-roughness	1 and 10 μm	Promote the procurement of bone cells and their mineralisation.
Nano-roughness	0.01 and 1 μm	Migration differentiation of osteoblasts, improves the osteointegration

2.8 Corrosion from titanium based dental implants

Corrosion may be general and localised. General corrosion observed uniform dissolution of the metal surface and localised corrosion distributed at specific sites on passive metal surface corresponding to high local dissolution (Williams & Chawla, 2014).