

Biomaterials in Implant Dentistry: An Overview

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Abstract

The objective of present day dentistry is to reestablish the patient to normal contour, function, comfort, esthetics, speech and health. What makes implant dentistry unique is the ability to achieve this goal regardless of the atrophy, disease, or injury of the stomatognathic system. However, the more teeth a patient is missing, the more challenging this task becomes. The biocompatibility profiles of synthetic substances (biomaterials) used for the replacement or augmentation of biological tissues have always been a critical concern within the health care disciplines. Therefore, the desire to positively influence tissue responses and to minimize biodegradation often places restrictions on which materials can be safely used within the oral and tissue environments.⁷ Improvements in both the quality and quantity of the implant biomaterial are the reasons for this treatment modality being practiced abundantly today. The development and modification of dental implants have

taken place in an effort to create an optimal interaction between the body and the implanted material. The goal of achieving an optimal bone-implant interface has been approached by the alteration of implant surface topography, chemistry, energy and charge as well as bulk material composition.³ This article reviews the various implant biomaterials and their suitability of use in implant dentistry.

Keywords: Biomaterials, Zirconium, Corrosion, Titanium, Hydroxyapatite, Bioceramics

Introduction

In the past 50 years, implant dentistry has evolved from an experimental treatment to a highly predictable treatment to replace missing teeth with implant-supported prostheses.¹ The implant has gained importance with the work of Professor Branemark, who studied microcirculation in bone tissue.² For many years, in an attempt to replace a missing tooth, many biomaterials have been evolved as implants in an effort to create an optimal interaction between the body and

the implanted material.³ Biomaterials are those materials which are compatible with the living tissues.⁴ These materials are mainly selected on the basis of their overall strength properties.⁴ Research on dental implant designs, materials and techniques has expanded in the past few years and it is expected to expand in the future due to the recent growth of the global market for dental implants and the rise in the demand for cosmetic dentistry.⁵ As a result of ongoing research in treatment planning, implant design, materials and techniques, unsurprising achievement is now a reality for the rehabilitation of many challenging situations.⁴

Requirements of Implant Biomaterials⁶: Any material planned for use in the fabrication of a dental implant must meet two essential criteria:

Firstly, the material must be chemically and biologically suited with the living tissue. That is, it must be **biocompatible**.

Secondly, the material must allow the implant design to be **bio functional** with respect to force transfer.

The biocompatible material must exhibit properties that enable it to be shaped into a configuration that takes optimal advantage of the available bone for implantation, while maintaining physical properties that meet the particular force requirements of a functioning implant.

Biocompatibility and **biofunctionability** are the two basic considerations in any discussion of biomaterial selection for fabrication of dental implants. If a material falls short in either of these regards, it is not fit for implant fabrication. The use of inappropriate biomaterial can compromise design in two ways:

First of all, the optimal use of available bone can be compromised by the use of a biomaterial which is mechanically weak.

Secondly, treatment protocol requirements may inhibit the use of a more ideal configuration design which are necessitated by the use of certain biomaterials. These two considerations are interrelated.

Important factors that influence biocompatibility are:

- Cytotoxicity of corrosion products
- Metal contamination
- Corrosion resistance

Classification of Implant Biomaterials

According to Misch⁷

Implant biomaterials are classified as:

1. Metals and Alloys
 - a. Titanium and Titanium-6 Aluminum-4 Vanadium (Ti-6Al-4V)
 - b. Cobalt-Chromium-Molybdenum-Based Alloys
 - c. Iron-Chromium-Nickel-Based Alloys
 - d. Other Metals and Alloys
2. Ceramics and Carbon
 - a. Aluminum, Titanium and Zirconium Oxides
 - b. Bioactive and Biodegradable Ceramics Based on Calcium Phosphates
3. Polymers and Composites
 - a. Polytetrafluoroethylene (PTFE)
 - b. Polyethylene terephthalate (PET)
 - c. Polymethylmethacrylate (PMMA)
 - d. Polyethylene (PE)
 - e. Polypropylene (PP)
 - f. Polysulfone (PSF)
 - g. Polydimethylsiloxane (PDS) or silicon rubber (SR)

According to O'Brien⁸

Implant biomaterials are classified as:

- A. Metals
 1. Cp titanium and Ti-6Al-4V
 2. Cobalt based alloy
- B. Ceramics

1. Inert ceramics
 - a. Alumina
 - b. Carbon
 - c. Zirconia
2. Bioactive ceramics
 - a. Bioactive glasses and glass ceramics
 - b. Calcium phosphates

According to Phillips⁹

Implant biomaterials are classified as:

A. Metallic Implants

1. Titanium
2. Titanium alloys (Ti6Al4V)
3. Commercially pure (CP)Ti
4. Stainless steel
5. Co-Cr-Mo alloys
6. Ni-Cr-Mo-Be alloy

B. Ceramic Implants

1. Alumina (Al₂O₃)
2. Zirconia (ZrO₃)
3. Ca₁₀(PO₄)₆(OH)₂
4. Hydroxyapatite (HA)
5. Tricalcium phosphate

C. Other implant materials

1. Carbon and carbon compounds (C and SiC)
2. Vitreous carbon

Metals and Alloys: Metals are probably the oldest form of material used for dental implants and are still by far the most common type of materials used today.¹¹

Metals are used as biomaterials due to their excellent electrical and thermal conductivity and mechanical properties.¹²

Titanium^{11,13:} Titanium was first discovered in England by William Gregor in 1791. It exists in nature as a pure element with an atomic number 22, with atomic weight 47.9. In nature, titanium exists mostly as

titanium dioxide (TiO₂) or imenite (FeTiO₃). The two forms of titanium used for endosseous dental implants are 'commercially pure' titanium (cpTi) and the titanium alloy Ti6Al4V. The most common titanium alloy, Ti6Al4V, supports a mixture of the alpha and beta phases at room temperature (25 °C).

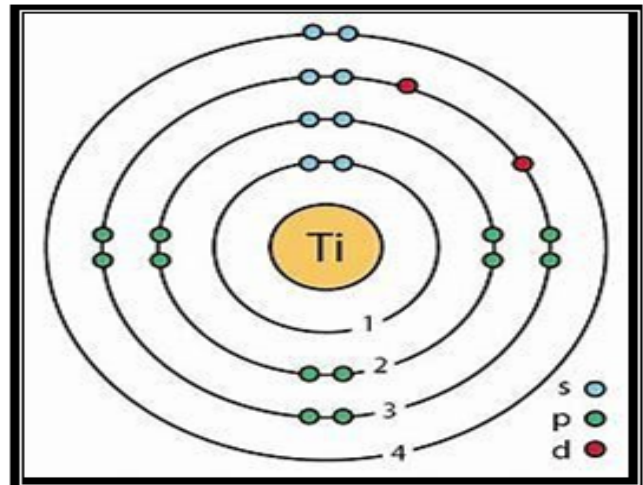


Figure 1: Atomic structure of Titanium

Titanium-6 Aluminum-4 Vanadium (Ti6Al4V): The main elements of the alloy are aluminum (5.5-6.5%) and vanadium (3.5-4.5%).¹² Titanium alloys exist in three forms alpha, beta and α - β . These types originate when pure titanium is heated with elements Al, Va in certain concentrations and cooled.¹⁴ The addition of alloying elements to titanium enables it to have a wide range of properties i.e. Aluminum tends to stabilize the α -phase, that is increase the transformation temperature from α - to β -phase. Vanadium stabilizes the β -phase by reducing the temperature of the transformation from α to β .¹² Titanium alloys, namely, Ti-6Al-4 V and Ti-6Al-7Nb, are used in extra low interstitial (ELI) grades. With the exception of pure titanium, the modulus of elasticity of Ti-6Al-4V is closer to that of bone than that of any other widely used metallic implant biomaterial. Newer titanium alloys have been developed, including Ti-13Nb-13Zr and Ti-15Mo-2.8Nb. These alloys utilize other phase stabilizers

instead of aluminum and vanadium, and they may exhibit greater strength and corrosion resistance.¹⁵

Commercially pure titanium (CPTi): Commercially pure titanium is also referred to as unalloyed titanium. American Society for Testing and Materials (ASTM) Committee F-4 on Materials for Surgical Implants recognizes four grades of commercially pure titanium i.e. commercially pure grade I titanium, commercially pure grade II titanium, commercially pure grade III titanium, and commercially pure grade IV titanium.¹⁶ It is an alloy of titanium & oxygen. To satisfy the British standard specification for use in surgical implants, the oxygen content must be less than 0.5%. In this form the alloy has a close packed hexagonal structure. Elements such as oxygen, nitrogen & carbon have a greater solubility in the close packed hexagonal structure of the alpha phase than in cubic form of beta phase. These elements form interstitial solid solutions with titanium & help to stabilise the alpha phase. Transition elements such as molybdenum, niobium & vanadium act as beta stabilisers.³

Cobalt-Chromium-Molybdenum-Based Alloy: The cobalt-chromium alloys were originally developed by Haynes in 1907 and were known as the Stellite alloys because of their bright metallic lustre.¹⁷ Cobalt-chromium-molybdenum cast alloys generally consists of 63% cobalt, 30% chromium and 5% molybdenum with small amounts of carbon, manganese and nickel. Molybdenum is a stabilizer; chromium provides the passivating effect to ensure corrosion resistance; and carbon serves as a hardener. American Society for Testing and Materials (ASTM) lists four types of CoCr alloys which are recommended for surgical implant applications: (1) cast CoCrMo alloy (F75), (2) wrought CoCrWNi alloy (F90), (3) wrought CoNiCrMo alloy (F562), and (4) wrought CoNiCrMoWFe alloy (F563).

Till date, only two of the four alloys are used mainly in implant fabrications, the castable CoCrMo and the wrought CoNiCrMo alloy.⁴

Iron-chromium-nickel-based alloys: This alloy is used mostly in a wrought and heat-treated metallurgical condition, which results in high-ductility and high-strength alloy. Since, nickel is a major element, its use should be avoided in patients allergic or hypersensitive to nickel. However, if a stainless-steel implant is modified prior to surgery, then recommended procedures call for repassivation to obtain an oxidized (passivated) surface condition to minimize in vivo biodegradation. The iron-based alloys have galvanic potentials and corrosion characteristics that could result in galvanic coupling and biocorrosion if interconnected with titanium, cobalt, zirconium, or carbon implant biomaterials.⁴

Stainless Steel: The first stainless steel for implant fabrication was the 18-8, which is stronger and more resistant to corrosion than the vanadium steel. Vanadium steel is not used now in implants as it is not resistant to corrosion in vivo. Later 18-8sMo stainless steel was introduced which contains a small percentage of molybdenum to improve the corrosion resistance in chloride solution (salt water). This alloy came to be known as type 316 stainless steel.¹² The specifications for surgical stainless steel, usually called 316 stainless steel, typically call for 17-20% chromium, 10-14% nickel, 2.5-4% molybdenum and a maximum of either 0.03 or 0.08% carbon depending on the application.¹⁷

Polymers¹⁸: Polymers have been utilized in medical implants for nearly 80 years. Polymethylmethacrylate (PMMA) was first used in the body in the 1930s and was initially chosen for its biocompatibility, stiffness and optical properties.¹⁸

Polyethylene (PE): PE was originally synthesized by a German chemist in 1898, Hans von Pechmann, as an accidental side-product during his investigations into diazomethane. It is a thermoplastic polymer. It becomes molten above a specific temperature and solidifies upon cooling in a reversible process and this can be repeated cyclically with no significant degradation. This is a compelling property driving widespread PE use, allowing for rapid moulding, injecting and manipulation into any shape and size desirable, from plastic packaging and shopping bags to construction materials such as drain pipes and high-performance devices including bulletproof vests and medical implants. Five major categories of PE exist, as defined by the American Society for Testing and Materials.

Low Density PE (LDPE) typically has lower tensile strength than other PE varieties but more desirable flow properties for the formation of plastic films and wraps which are used in plastic shopping bags and plastic wrap, as well as sterile medical packaging and films.

Linear Low-Density PE (LLDPE) is a more linear form of LDPE and therefore has shorter branches and higher tensile strength. LLDPE is a popular choice for a wide range of products including toys, containers, pipes and bubble wrap.

Medium Density PE (MDPE) can be produced from a range of catalysts including Ziegler-Natta as well as chromium/silica and metallocene. With lower density and better stress cracking resistance than the high-density equivalents, MDPE has more functional uses in plumbing and piping, with limited medical applications.

Cross-linked PE (XPE) is not defined by a specific molecular weight range but refers to an analogue of PE. XPE is a cross-linked equivalent of MDPE or HDPE with intermediate strength. Primary applications of XPE include replacing copper as a plumbing material as well

as an insulator for use in medium-to high-voltage cables.

High density PE (HDPE): features a high density-to-strength ratio and tensile strength, but exhibits a low melting point and tendency to deform under high temperature and pressure, necessary for autoclave sterilization for medical applications. It is commonly used in medical implant devices. In addition to all the properties of HDPE, **ultra-high molecular weight PE (UHMWPE)** features extremely long polymer chains which provide material properties suitable for high-load applications. It is widely used and typically has a molecular weight of about 4-6 million g/mol, an elastic modulus on the order of 1 GPa and crystallinities of about 50–60%.¹⁸ UHMWPE has diverse applications when formed into fibres, including for personal and vehicle armour, suspension lines and cables.

Polymethyl Methacrylate (PMMA): Poly methyl methacrylate (PMMA) is an acrylic resin usually used with a long tradition for prosthetic purposes.¹⁹ They are commonly employed in orthopedics and dentistry as the grouting material between bone and a synthetic implant.¹⁸ PMMA is used broadly in medical applications such as a blood pump and reservoir, an IV system, membranes for blood dialyzer and in in vitro diagnostics.¹²

Polypropylenes (PP): Polypropylene is one of the biocompatible, biostable polymer used widely in clinical applications ranging from sutures to load-bearing implants.²⁰ They are commonly used in finger joint prostheses, grafts and sutures. Typically, the molecular weight ranges between 200,000– 700,000 g/mol and the density is generally bounded between 0.85-0.98 g/cc. This polymer material has an exceptional fatigue life in flexion and it is for this reason that PP is used in small joint reconstruction. The

high tensile strength of this polymer also serves as a good material choice for non-resorbable sutures.¹⁸

Polystyrene (PS) And Its Co-Polymers: PS is polymerized by free radical polymerization and is usually atactic. Three grades are available: unmodified general purpose PS (GPPS, Tg: 100 °C), high impact PS (HIPS) and PS foam. GPPS has good transparency, lack of color, ease of fabrication, thermal stability, low specific gravity (1.04 ~1.12 g/cm³) and relatively high modulus. PS is mainly processed by injection molding at 180~250 °C. To improve processability additives such as stabilizers, lubricants and mold-releasing agents are formulated.¹²

Polyamides (NYLONS): Polyamides are known as nylons and are designated by the number of carbon atoms in the repeating units. They have excellent fiber-forming ability due to interchain hydrogen bonding and a high degree of crystallinity, which increases strength in the fiber direction. Nylons are hygroscopic and lose their strength in vivo when implanted.¹²

Polyvinylchloride (PVC): PVC is an amorphous, rigid polymer due to the large side group (Cl, chloride) with a Tg of 75~105 °C. It has a high melt viscosity; hence it is difficult to process. To prevent the thermal degradation of the polymer (HCl could be released), thermal stabilizers such as metallic soaps or salts are incorporated. Lubricants are formulated on PVC compounds to prevent adhesion to metal surfaces and facilitate the melt flow during processing.¹²

Silicon Rubber (SR) or Polydimethylsiloxane (PDMS): Silicones are a broad class of polymers also known as siloxanes.¹⁸ Silicone rubber, developed by Dow Corning company, is one of the few polymers developed for medical use. Low molecular weight polymers have low viscosity and can be cross-linked to make a higher molecular weight, rubber-like material.

Medical grade silicone rubbers contain stannous octate as a catalyst and can be mixed with a base polymer at the time of implant fabrication.¹²

Bioceramics: Bioceramics have structural function as joint or tissue replacement. It can be used as coatings to improve the biocompatibility of metal implants and as resorbable lattices. The thermal and chemical stability of ceramics, high strength, resistance to wear and durability makes a ceramic good material for surgical implants.²¹ Bioceramics are non-toxic. They can be bioinert, bioactive that is durable material which can undergo interfacial interactions with surrounding tissue, they can be also biodegradable, soluble or resorbable.²¹

Alumina: Alumina was firstly used as implant material in 1970s.²¹ Aluminum oxide is very hard (9 on the Mohr scale), stable to high temperatures and chemically durable. Polycrystalline sintered alumina is very strong compared to most ceramics, although it is brittle. Its strength, abrasion resistance and chemical inertness make it attractive for dental and bone implants. Very pure alumina sinters only with difficulty at high temperatures (~1900 °C). Addition of a small amount of magnesia gives a material with many fewer pores; commercial "Lucalox" is made this way and is translucent, so that it can be used as an arc tube in sodium vapor lamps. If the starting material contains a small amount of alkali (sodium or potassium) oxide, liquid forms during sintering, so that densification takes places more rapidly and at lower temperatures. However, the liquid forms a glassy phase at the grain boundaries, making the alumina weaker at high temperatures and less chemically durable.²²

Zirconia (ZrO₂): Zirconium was discovered in 1789 by the German chemist M.H. Klaproth.²³ Zirconia (zirconium dioxide, ZrO₂) have three different crystal lattices (phases), depending on temperature: monoclinic

(m), from room temperature to about 1100; tetragonal (t), from 1100 °C to about 2400 °C; and cubic (c), above this temperature. Due to the different shapes of the crystals, the three phases have different densities.²⁴

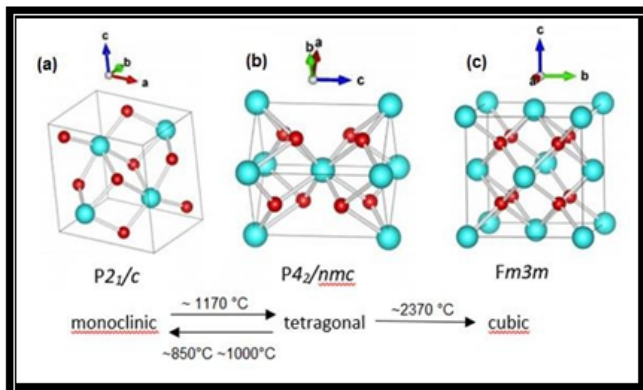


Figure 2: Phase diagram of Zirconia

Three types of zirconia-containing ceramic materials used for dental application are:

- Yttrium stabilized zirconia polycrystals (tetragonal zirconia polycrystals 3Y-TZP).
- Transformation toughened partially stabilized zirconia with magnesium (Mg-PSZ), and
- Dispersion toughened ceramics - zirconia toughened alumina (ZTA).

Yttrium Stabilized Zirconia Polycrystals: It is a bioinert material with high mechanical properties and is 6 times harder than stainless steel.²³ It has a very small grain size, extremely low porosity and a unique property referred to as “transformation toughening.” This transformation is associated with 3-5% volume expansion, which induces compressive stresses, thereby closing the crack tip and preventing further crack propagation. This mechanism is known as transformation toughening and confers Y-TZP with superior fracture strength and fracture toughness compared with other dental ceramics.²⁵

Glass-Infiltrated Zirconia Toughened Alumina (ZTA): Zirconia-based ceramics based can be combined

with a matrix of alumina (Al₂O₃) to favourably utilize the stress-induced transformation ability of zirconia and produce a structure known as ZTA (alumina reinforced with zirconia grains). ZTA is developed by addition of 33 vol% of 12 mol% ceria-stabilized zirconia (12Ce-TZP) to In-Ceram alumina. Slip-casting or soft machining is utilized to process In-Ceram zirconia and then it is glass infiltrated. As compared to 3Y-TZP dental ceramics, ZTA exhibits lower mechanical properties due to incorporation of residual porosity (8-11%).²⁵

Alumina Toughened Zirconia (ATZ): ATZ is a composite ceramic material combining of 20 wt% alumina and 80 wt% zirconia containing 3 mol% yttria. Resistance to surface degradation (LTD) of the material has been improved by the addition of alumina as low as 0.25 wt% to TZP ceramics significantly improves the resistance of the material to surface degradation (LTD). ATZ exhibits the highest bending strength known for ceramics, both at room temperature (1,800-2,400 MPa) and at elevated temperatures (>800 MPa at 1000 °C) and this property results in a high thermal shock resistance of ATZ.²⁵

Calcium Phosphate Ceramics: Calcium phosphate-based bioceramics have been in use in medicine and dentistry for nearly 20 years.²⁶ Applications include dental implants, percutaneous devices, use in periodontal treatment, alveolar ridge augmentation, orthopedics, maxillofacial surgery, otolaryngology and spinal surgery.²⁶ Calcium-phosphate-based bioceramics have also been used as coatings on dense implants and porous surface layers to accelerate and enhance fixation of a substrate biomaterial to tissue.²⁷

Hydroxyapatite: Hydroxyapatite is a bioactive material which is used in many prosthetic applications. Clinical tests have proved that it is compatible with the

tissues of vertebrates, thus making it an attractive bioimplant material.²⁸



Figure 3: Hydroxyapatite coated implants

Coatings of hydroxyapatite are often applied to metallic implants to alter the surface properties. To date, the only commercially accepted method of applying hydroxyapatite coatings to metallic implants is plasma spraying.

Bioglass/Glass Ceramic: Glass ceramics were developed by S. D. Stookey of Corning Glass Works in the early 1960s.¹² Since discovery of the bioglasses, which bond to living tissue by Hench and Wilson, various kinds of bioactive glasses and glass ceramics with different functions such as high mechanical strength and fast setting ability have been developed.²¹ Bioactive glasses mainly consist of four fundamental components namely, silicon oxide, sodium oxide, calcium oxide and phosphorus pentoxide.²⁹

Aluminum, Titanium And Zirconium Oxides: The aluminum, titanium and zirconium oxide ceramics have a clear, white, cream, or light-gray color, which is beneficial for applications such as anterior root form devices. In early studies of dental and orthopedic devices in laboratory animals and humans, ceramics have exhibited direct interfaces with bone, similar to an osseointegrated condition with titanium. In addition, characterization of gingival attachment zones

along sapphire root form devices in laboratory animal models has demonstrated regions of localized bonding.

Carbon And Carbon Silicon Compounds: Carbon compounds are often classified as ceramics because of their chemical inertness and absence of ductility.⁷ Carbons can be made in many allotropic forms: crystalline diamond, graphite, noncrystalline glassy carbon and quasicrystalline pyrolytic carbon. Out of these, only pyrolytic carbon is widely utilized for implant fabrication. It is normally used as a surface coating and is also used to coat surfaces with diamond. The techniques of coating with diamond have the potential to revolutionize medical device manufacturing, still they are not yet commercially available.¹²

Factors Affecting Implant Biomaterials

Mechanical Factors³

Implant Configuration

Implant Diameter: Implant diameter is the measurement measured from the peak of the largest thread to a similar point on the inverse side of the implant. As of now implants vary in diameter from 3 to 7 mm. The necessities for implant diameter depend on both surgical and prosthetic prerequisites. Width of implant is designed to gain maximum stability. Distribution of the stress depends on diameter of the implant and distribution of stress varies with respect to implant design.

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Implant Length: It is the length from the platform to the peak of implant. Length of the implant was

delegated short, medium and long. The scope of short implant length was from 6-9 mm. The medium implant length lies between 10-12 mm and long implant length ranges from 13-18 mm. The 7 mm implant length prompts to more dissatisfaction rate while comparing with other implant lengths.

Implant shape: Most current implant frameworks are accessible as strong or hollow screws or cylinders. Among screw shaped implants, extensive modification has been made to the crestal and apical bit of the implant to expand self-tapping. These stepped cylindrical implants demonstrate more even stress dispersal contrasted with tube shaped or decreased implants. The stability of the implant at the time of placement is dependent upon bone quantity and quality. According to Lekholm and Zarb, the bones can be classified as type D1, D2, D3 and D4.³¹

Chemical Factors

Corrosion^{32,33}: Corrosion is defined as the process of material degradation due to electrochemical oxidation because of the environment-material interaction. Metals undergo corrosion when exposed to an aqueous environment of the body fluids and involves the formation of localized electrochemical cells. Different types of corrosion of dental materials are:

Galvanic Corrosion: It is defined as dissolution of metals caused by macroscopic differences in electrochemical potentials, usually because of dissimilar metals in proximity. It is caused by improper use of metals, e.g., a stainless-steel cerclage wire in contact with a cobalt or titanium-alloy femoral stem, a cobalt-alloy femoral head in contact with a titanium-alloy femoral stem, and a titanium-alloy screw in contact with a stainless-steel plate.

Pitting and Crevice Corrosion: Pitting is a form of localized, symmetric corrosion in which pits are formed locally on the metal surface. Crevice is a type of localized corrosion due to differences in oxygen pressure or concentration of electrolytes or changes in pH within a confined space, such as in the crevices between a screw and a plate.

Stress Corrosion Cracking: It is a phenomenon in which a metal, when exposed to a certain environment, particularly those rich in chlorides, is subjected to stress and fails at a much lower level of stress than usual, due to corrosion. Pitting and cracking due to stress corrosion, while generally associated with stainless steel in chloride media, have not been observed on recovered surgical implants.

Surface Specific Properties³⁴

Hydroxyapatite³⁵: HAP is a ceramic biomaterial that has been found to exhibit good bioactivity, biocompatibility, osteoconductivity, nontoxicity, and a non-inflammatory nature under both in vitro and in vivo conditions. But due to its highly brittle nature, HAP cannot be used alone as an implant in load-bearing applications. Thus, load-bearing implants can be coated with HAP. Because of high biocompatibility with bone, connective tissues, and epithelium, the HAP-coated implants have attracted attention for dental implant applications.

Hydroxyapatite Coating Methods

Titanium Plasma Spraying^{7,34}: Plasma spraying is a typical additive modification for titanium surfaces and can produce a rough surface usually by the deposition of HA. In this approach, HA ceramic particles are injected into a plasma torch at high temperatures. The particles are projected onto the titanium surface and allowed to condense and fuse

together. Kay et al. used scanning electron microscopy (SEM) and spectrographic analyses to show that the plasma-sprayed HA coating could be crystalline and chemically and mechanically compatible with dental implant applications.

Hot isostatic pressing technique³⁵: Hot isostatic pressing (HIP) is a metallurgical process employed to decrease the porosity and increase the density of ceramic materials. The method is used to improve mechanical properties and workability of the materials. HIP offers a method to produce dense HAP coating on a Ti substrate. In this technique, the pressure is exerted using gas at a high temperature. A dense coating with HAP structure with little porosity was obtained using hot isostatic pressing for 35 min at 700–850°C, and a maximum pressure of 1000 bar.

Pulsed laser deposition coating technique³⁵: In this technique, a high-power pulsed laser beam is focused inside a vacuum chamber to strike a target of the material that is to be deposited. This method is simple and has the advantages of forming a coating with a uniform thickness on flat substrates, the ability to control deposition parameters, and the ability to produce high crystalline coatings.

Electrophoretic deposition coating technique: This is a coating process based on the electrode deposition in which colloidal particles suspended in a liquid migrate under the influence of an electric field (electrophoresis) and are deposited onto an electrode. The method can be applied for ceramic coatings, and has the advantages of cost-effectiveness, and flexibility, as it can be used to coat objects with a complex shape and morphology. Electrophoretic deposition coating is particularly helpful in the deposition of HAP coatings because this technique

permits the control of the composition, thickness, and microstructure of coatings.

Thermal spraying technique: This is a coating method in which melted or heated materials are sprayed onto a surface. This technique is a collection of different coating methods, which provides functional surfaces to defend or increase the function, and it can be categorized into three major classes, namely, electrical arc spray, plasma arc spray, and flame spray.

Dip coating techniques: In the dip coating method, the substrate is dipped and withdrawn from the solution. This coating consists of three major steps, namely, dipping, withdrawing, and drying, and curing or sintering. After withdrawing the substrate from the medium, a consistent liquid film is carried on the substrate, and after evaporation, a thin coat of deposits is made. In most cases, after drying, an extra post-treatment stage, such as sintering or curing, is desired to acquire the final coating. This method has many advantages, such as being inexpensive, fast, simple, and it offers uniformity of coating and the ability to coat irregular shapes and patterns.

Sol-gel technique: This method is widely used in the synthesis of many inorganic materials because of the easy development of crystalline homogeneous films at relatively low temperatures. Further, this method can also be conveniently used for making complex shape coatings. In this method, generally, two solvents are mixed with a calcium phosphate (CaP) precursor. The most used solvents for the sol preparation are water and ethanol. The sol-gel method is a low cost and simple method, which permits the molecular level mixing of the HAP precursors.

Sputter coating technique: This method is a vapor deposition technique for making a thin coating by

sputtering, and it consists of ejecting a substance from the target to a substrate using energetic particle bombardment. In this method, a gas plasma such as neon, argon, krypton, or xenon is used to eliminate substances from a negatively charged target, which are then deposited as a thin film on the substrate.

Anodization: Titanium surfaces can be modified by potentiostatic or galvanostatic anodization in strong acids such as H₂SO₄, H₃PO₄, HF, HNO₃ at high current density or potential. The result is an anodized surface with a thicker oxide layer (usually more than 100 nm) that nevertheless appears to be mostly minimally rough. In certain respects, increased oxide layer thickness by anodization can be considered an additive modification process.

Grit Blasting: Blasting the titanium surface, with either titanium oxide or alumina particles, is another method by which surface roughness can be increased. Generally, the particles are projected through a nozzle at high velocity (usually by compressed air). Roughing the surface by blasting generally produces a surface that is moderately rough.

Acid Etching: Even though titanium is corrosion resistant, some acids can still be used to etch the surface. Roughening the surface by this method usually involves etching with strong acids such as HF, HNO₃, H₂SO₄ or HCl. Etching will remove small amounts of material to create micro pits on the surface. It can also reduce the carbon content in the oxide layer. Etched surfaces tend to be minimally rough with many displaying Sa values of $\leq 1 \mu\text{m}$. Apart from altering the topography of titanium post-etching, acid etching can also chemically modify the titanium surface.

Acid Etching Followed By Grit Blasting: The combination of blasting and etching is currently being used routinely for surface modification. This technique

involves the blasting of titanium surfaces with either alumina or titanium particles, followed by acid etching. Surfaces created by blasting and etching are generally moderately rough. The main rationale for combining these two surface treatments was to create a surface with optimal roughness for mechanical fixation, including the formation of smoothed sharp peaks that add a high frequency component on the surface.

Recent Advances in Implant Biomaterials: Recent developments related to titanium-based implant biomaterials²⁵: Recent trends in the research of titanium-based biomaterials show the aim to develop alloys composed of nontoxic and nonallergenic elements with excellent mechanical properties (low modulus and high strength) and workability. A new alloy for manufacturing narrow diameter implants (Roxolid, Straumann, Basel, Switzerland) has recently been introduced in dentistry. This alloy is based on the binary formulation of 83-87% titanium and 13-17% zirconium. It has been claimed that this alloy exhibits better mechanical characteristics compared to CpTi and Ti-6Al-4V, with a tensile strength value of 953 MPa and a 40% higher fatigue strength. The addition of zirconia to titanium leads to improved osteointegration and the alloy of zirconia and titanium is more biocompatible as compared to pure titanium. Another promising α -titanium alloy used as a surgical implant material is Ti_{12.5}Zr_{2.5}Nb_{2.5}Ta (TZNT). This alloy had a unique advantage of having a modulus of elasticity closer to the human bone (100 Gpa) as compared to the conventional titanium alloys. (120 Gpa) and it also possesses approximately equivalent admission strain (0.65%) with that of human bones (0.67%). The addition of elements such as Zr, Nb, and Ta to the alloy have shown no toxicity or any adverse tissue reactions and displays a better corrosion resistance.²⁵

Poly-ether-ether-ketone (PEEK)^{36,37}: Following confirmation of its biocompatibility two decades ago, polyaryletherketone polymers (PAEKs) have been increasingly employed as biomaterials for orthopedic, trauma, and spinal implants. Polyaryletheretherketone, commonly referred to as PEEK, is a member of the PAEK polymer family that has been used for orthopedic and spinal implants. PEEK is a high performance semi-crystalline thermoplastic polymer, which combines its very good strength and stiffness with an outstanding thermal and chemical resistance - e.g., against oils and acids. Being colorless and endowed with an elastic modulus close to that of the bone, PEEK is a viable option for dental implant manufacturing.

Conclusion

Implant restorative treatments in dentistry provide a significant opportunity to better understand the roles of biomaterial and biomechanical properties and their relationship to biocompatibility criteria for all types of surgical implants. Various biomaterials and designs are being used for long term treatment modalities.³⁸ Surface characterization and working knowledge about how surface and bulk biomaterial properties interrelate to dental implant biocompatibility profiles represent an important area in implant-based reconstructive surgery.⁷ Existing science and technology support the need for mechanically and chemically anisotropic substances to replace functional, load-bearing tissues and afford a most promising future for dental implant research and development. Multidisciplinary analyses should provide the basis for quantitative classifications of interfacial phenomena, and, thereby, the directly associated clinical longevities.³⁸

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