



DENTAL IMPLANT BIOMATERIALS – AN OVERVIEW

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ABSTRACT

Appropriate selection of the implant biomaterial is a key factor for long term success of implants. The biologic environment does not accept completely any material so to optimize biologic performance, implants should be selected to reduce the negative biologic response while maintaining adequate function. Every clinician should always gain a thorough knowledge about the different biomaterials used for the dental implants. This article makes an effort to summarize various dental bio-materials which were used in the past and as well as the latest.

Key words:

Dental Implants, Biomaterials

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INTRODUCTION

Implants have been gaining popularity amongst the patients and frequently are being considered as a first treatment option. Modern dentistry is beginning to understand, realize, and utilize the benefits of biotechnology in health care. Study of material sciences along with the biomechanical sciences provides optimization of design and material concepts for surgical implants. With all the advancements and developments in the science and technology, the materials available for dental implants also improved. The choice of material for a particular implant application will generally be a compromise to meet many different required properties. There is, however, one aspect that is always of prime importance that how the tissue at the implant site responds to the biochemical disturbance that a foreign material presents. The development and modification of dental implants have taken place for many years 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. Schmidt *et al.* (2001) defines an ideal bone implant material as having a biocompatible chemical composition to avoid adverse tissue reaction, excellent corrosion resistance in the physiologic limits, acceptable strength, a high resistance to wear and a modulus of elasticity similar to that of bone to minimize bone resorption around the implant.¹

Dental implant materials encompass a variety of materials with different degrees of interaction with the human body. Immediately after insertion of the implant, this contact

provides primary stability through friction and mechanical interlocking between the implant thread surface and the bone trabeculae. During the following weeks, the peri-implant bone is remodelled and replaced by newly formed bone. The major part of the final bone-implant contact is thus based on newly formed bone that originates from the adjacent peri-implant bone and is laid down on the implant surface in an osteoconductive manner.^{3,4}

Classification

Bioinert: The term bioinert refers to any material that once placed in the human body has minimal interaction with its surrounding tissue, examples of these are titanium, Zirconium, alumina.

Bioactive: Bioactive refers to a material, which upon being placed within the human body interacts with the surrounding bone and in some cases, even soft tissue. Prime examples of these materials are synthetic hydroxyapatite, glass ceramic and bioglass.

Bioresorbable: It refers to a material that upon placement within the human body starts to dissolve (resorbed) and slowly replaced by advancing tissue (such as bone). Common examples of bioresorbable materials are tricalcium phosphate [Ca (PO)] and ^{2,3,4} polylactic, polyglycolic acid, copolymers. Calcium oxide, calcium carbonate and gypsum are other common materials that have been utilised during the last three decades.⁵

Types of Biomaterials

Dental Implant Materials

Metals and Alloys (Titanium & Titanium –6 Aluminum4Vanadium (Ti6Al4V) and cp Ti, Cobalt Chromium Molybdenum Based Alloy, Iron Chromium NickelBased

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Alloys), Ceramics (Aluminum, Titanium and Zirconium oxide, Bioactive and biodegradable ceramics) Carbon Carbon & carbon silicon, Vitreous and Pyrolytic) Polymers and Composites (Poly methyl methacrylate (PMMA), Polyethylene (UHMWPE), Polytetrafluoroethylene (PTFE), Silicone rubber, Polysulfone)¹³

Bone Augmentation Materials

CERAMICS (Calcium phosphate, Bioactive glass & glass ceramics), POLYMERS (PMMA, Lactic/glycolic acid), NATURAL MINERALS (Collagen, Demineralized bone matrix, Bone morphogenic proteins)¹³.

Titanium and Titanium alloys Ti6Al4V

Titanium exists in nature as a pure element with an atomic number 22, with atomic weight 47.9. Titanium makes up about 0.6% of earth's crust and is a million times more abundant than gold. This metal exists as Rutile (TiO₂) or Ilmenite (FeTiO₃) compounds and requires specific extraction methods to be recovered in its elemental state.

Commercially Pure Titanium

Commercially pure titanium (Ti CP) and extra low interstitial Ti-6Al-4V (ELI) are the two most common titanium base implant biomaterials. These materials are classified as biologically inert biomaterials. As such, they remain essentially unchanged when implanted into human bodies. The human body is able to recognize these materials as foreign and tries to isolate them by encasing it in fibrous tissues. However, they do not promote any adverse reactions and are tolerated well by the human tissues. Its very good biocompatibility is due to the formation of an oxide film (TiO₂) over its surface. This oxide is a strong and stable layer that grows spontaneously in contact with air and prevents the diffusion of the oxygen from the environment providing corrosion resistance. It is a biomaterial with a high superficial energy and after implantation it provides a favourable body reaction that leads to direct apposition of minerals on the bone-titanium interface and titanium osseointegration (Acero *et al.*, 1999).

Ti6Al4V alloy

Ti6Al4V alloy is widely used to manufacture implants. The addition of alloying elements to titanium enables it to have a wide range of properties because aluminium tends to stabilize the alpha phase and vanadium tends to stabilize the beta phase, lowering the temperature of the transformation from alpha to beta. The alpha phase promotes good weldability, excellent strength characteristics and oxidation resistance. The addition of controlled amounts of vanadium as a beta stabilizer causes the higher strength of beta-phase to persist below the transformation temperature which results in a two-phase system. The beta phase can precipitate by an ageing heat treatment.

Low Modulus Titanium alloys

The Ti6Al4V alloy has some disadvantages: its elastic modulus, although low, is 4 to 6 times that of cortical bone and has low wear resistance that is a problem in articulations surfaces. Also, V can cause potential cytotoxicity and adverse tissue reactions (Steinemann, 1980), and Al ions from the alloy might cause long-term Alzheimer diseases (Rao *et al.*, 1996). Briefly, a biocompatible titanium base alloy suitable for bone

implant should meet at least the following requirements (Mehta, 2008):

- ✓ Potentially toxic elements, such as vanadium, copper and tin, should be avoided completely
- ✓ Elements that may have potential toxicological problems, such as chromium, nickel and aluminium, should be used only in minimum, acceptable amounts
- ✓ The alloy should have high corrosion resistance
- ✓ The alloy should have, at least, the following desirable mechanical properties: low modulus, high strength and good smooth and notched fatigue strength
- ✓ The alloy should have good workability and ductility.

Titanium-Hydroxyapatite Composite

Biocomposite materials have been developed in order to combine bioactivity of ceramics and mechanical properties of metals. Hydroxyapatite (HA) is known for its weakness and brittles but has an excellent biocompatibility and is a bioactive material. When HA is added to titanium, an improvement of the biomaterial chemical properties occurs. New developments try to aggregate hydroxyapatite as a second phase to the Ti alloy, with powder metallurgy techniques (P/M). In this composite material, particles of HA are incorporated in a porous titanium matrix providing points of good bone reaction. These solutions allow improved adhesion strength of the load bearing metallic component to the bone, resulting in shorter healing periods as well as predictable behaviour of the implant for longer periods of time. Although this benefits, there are some problems in the manufacture of the composite material and some doubts about its biocompatibility.

Ceramics

Implant research has focused on discovering tooth colored implant material that improves the aesthetic appearance of dental implants and at the same time is highly biocompatible and able to withstand the forces present in oral cavity. Ceramic implants can withstand only relative low tensile or shear stress induced by occlusal loads, but they can tolerate quite high levels of compressive stress. Ceramics can either be plasma sprayed or coated on to the metallic surfaces which can be more thermodynamically stable, hydrophilic, and non conductive of heat and electricity, thereby producing a high strength integration with bone⁵. Aluminium oxide (Al₂O₃) is used as a standard biomaterial for ceramic implants because of its inertness (Biostability) with no evidence of adverse in vivo reactions. Zirconia (ZrO₂) has also demonstrated a high degree of inertness, although alumina has higher surface wettability compared with other surfaces, such as those of metallic implants. These types of ceramic implants are not bioactive in that they do not promote the formation of bone⁵.

Alumina

High-density, high-purity (99.5%) Al₂O₃ was chosen for dental implant manufacturing because of its combination of excellent corrosion resistance, good compatibility, high wear resistance, and high strength. Although some dental implants were made of single-crystal sapphire characterized by a glassy appearance, major interest in literature was shown for fine grained polycrystalline α -Alumina (α -Al₂O₃) produced by pressing and sintering at temperatures ranging from 1600 to 1800 °C (depending upon the properties of the raw material). When preparing Alumina bulk material, to achieve a fully

dense sintered body with a fine grain microstructure, a very small amount of MgO (0.5 %) should be used as a grain growth inhibitor. Likewise, the amount of SiO₂ and alkali oxides ought to be kept below 0.1 %, as they impede densification and promote grain growth. Also the amount of CaO has to be below 0.1 %, since its presence lead to the lowering of the static fatigue resistance. Anyway, the high hardness and modulus of elasticity make the material brittle. Combined with the relatively low bending strength and fracture toughness, the material is prone to fracture when loaded unfavorably. This might be the reason why dentists do not use Alumina implants. Interestingly, however, fracture was seldom mentioned in the literature as a reason for implant loss. The overall survival rate of Alumina implants was lower than that of Titanium implants as assessed in a series of systematic reviews where 95.4 % of the implant supported single crowns and 96.8 % of the implant supported fixed partial dentures survived at 5 years . The only exception where long-term survival rates with Alumina implants were comparable to Titanium implants is the investigations by Fartash *et al.*

Zirconia

The name of the metal zirconium originates from the Arabic “zargun” (golden in color), which in turn comes from the two Persian words Zar (gold) and Gun (color). Zirconia, the metal dioxide (ZrO₂), was identified as such in 1789 by the German chemist Martin Heinrich Klaproth in the reaction product obtained after heating some gems. Its mechanical properties are close to those of metal. In 1975, Garvie *et al.* proposed a model to rationalize the good mechanical properties of Zirconia, by virtue of which it has been called “ceramic steel”. At ambient pressure, unalloyed Zirconia can assume three crystallographic forms depending on the temperature. At room temperature and upon heating up to 1170 ° C, the symmetry is monoclinic (P21/c). The structure is tetragonal (P42/nmc) between 1170 and 2370 °C and cubic (Fm3m) above 2370 °C and up to the melting point. The transformation from the tetragonal (t) to the monoclinic (m) phase upon cooling is accompanied by a substantial increase in volume (*4.5 %), sufficient to lead to catastrophic failure. This transformation is reversible and begins at *950°C on cooling. Alloying pure Zirconia with stabilizing oxides such as CaO, MgO, Y₂O₃, or CeO₂ allows the retention of the tetragonal structure at room temperature and therefore the control of the stress-induced transformation, efficiently arresting crack propagation and leading to high toughness. Anyway tetragonal form cannot withstand more stress. When a crack develops, tetragonal grains convert immediately to monoclinic form. As the crack propagates, sufficient stress develops within the tetragonal structure and the grains around the crack transform to stable monoclinic form. In this process, expansion volume of zirconium dioxide crystals occurs, which produces compressive stress around the crack preventing further propagation. This mechanism is known as Transformation toughening and is influenced by temperature, vapor, particle size, micro- and macrostructure, and concentration of stabilizing oxides.

As already stated, Zirconia showed a high level of biocompatibility. In vitro experiments on different cell lines, in vivo studies on animals, and clinical studies on humans proved the safety of this material. Evidence from in vitro studies maintained the osteo conductivity of Zirconia ceramics. In their preliminary in vitro investigation, Kohal *et al.*, concluded

that one-piece Zirconia implants restored with all ceramic crowns possibly fulfilled the biomechanical requirements for anterior teeth. In another study, mean fracture strength of Zirconia implants was investigated after chewing simulation and found to be within the limits of clinical acceptance. However, preparation of a one-piece Zirconia implant to receive a prosthesis significantly compromised fracture strength. The authors concluded that long-term clinical data were necessary before one-piece Zirconia implants could be recommended for clinical practice. Two-piece Zirconia implants were considered clinically inadequate owing to the increased risk of fracture at the level of the implant head . When implanted in bone or soft tissues, the latter react favorably with undetectable residue release and almost no fibrous encapsulation and inflammatory reactions are observed . An animal study conducted by Scarano *et al.* found that unloaded Zirconia implants osseointegrate when inserted in rabbit's tibia bones without any signs of inflammation or mobility. Sennerby *et al.* reported superior osseointegration of modified (roughened) Zirconia implants when compared to machined ones and similar resistance to removal when compared to oxidized Titanium implants. Loaded Zirconia implants were studied and compared to Titanium ones by Kohal *et al.* who concluded that there was no difference as for osseointegration level between the two groups. In contrast, Akagwa *et al.* reported evident crestal bone loss around loaded Zirconia implants when compared to an unloaded group. Yet, bone-implant contact of the two groups was similar in this study. Five years later, the same research group reported possible long-term and stable osseointegration of loaded and unloaded Zirconia implants.

Glass Ceramics

They are bioactive ceramics first introduced in 1971. Bioglass or Ceravital Silica based glass with additions of calcium and phosphate produced by controlled crystallization. It has high mechanical strength, less resistant to tensile and bending stresses, extremely brittle and they chemically bond to the bone due to formation of calcium phosphate surface layer.

Hydroxyapatite

It was successfully used as implant material in 1988 at North America and to begin with for repair of residual ridge resorption in 1970s. It is similar to the mineral component of bones and hard tissues in mammals. This material has capability to integrate in bone structure and support in growth of the bone. It is thermally unstable with low mechanical strength to withstand long term load bearing applications.

Plasma sprayed hydroxyapatite was first used by Herman in 1988 . Crystalline HApowder is heated to a temperature of 12000 to 16000 °C in a plasma flame formed by a electric arc through which an argon gas stream passes. HA particle size is approximately 0.04mm . The particles melt and are sprayed on to the substrate, they fall as drops and solidify. Round interconnected pores are formed. Coating bonds to substrate by mechanical interlocking. There is a lot of controversy regarding the ideal coating thickness of HA coating. Studies have shown that fracture occurred in coatings which were more than 0.1mm in thickness whereas bioresorption was unacceptably rapid with coatings less than 0.03mm. Ideal coating thickness of 0.05 mm is recommended .^{6,7}

Bioresorbable

Calcium Phosphate

These ceramics have biochemical composition similar to natural bone and form direct chemical bonding with surrounding bone due to presence of free calcium and phosphate compounds as implant surface. It has excellent biocompatibility, no local or systemic toxicity, no alteration to natural mineralization process of bone, lower mechanical tensile, shear and fatigue strength. It is brittle with low ductility. It exists in dense or porous form. The pores though decrease the strength they increase the surface area providing additional region for tissue in growth. Ideal pore size is around 150µm, same diameter as shown by inter trabecular spaces in bone. Calcium Phosphate Ceramics show varied degree of resorption or solubility in physiologic fluids. The resorption depends on Crystallinity. High crystallinity is more resistant to resorption. Large particles size requires longer time to resorb. Greater the porosity, more rapid is the resorption. Resorption is more at low pH eg. in case of infection or inflammation. Presence of impurities accelerate resorption. It has been seen that HA resorb less readily than Tri Calcium Phosphate.

Polymers

Polymeric implants were first introduced in 1930s. However they have not found extensive use in implant due to low mechanical strength and lack of osseointegration. Poly-ether-ether-ketone (PEEK) Ceramic dental implants made of Zirconia seem to be a better suitable alternative to Titanium because of its tooth like color, mechanical properties, biocompatibility, and low plaque affinity. However, at the moment, the absence of a scientific consensus and the poor market share of Y-TZP may challenge the above statement. Interestingly, the systematic review of the literature by Andreiotelli already cited (published in 2009) concluded that the scientific clinical data were not yet sufficient to recommend ceramic implants for routine clinical use. This corresponds to a grade C recommendation of the definitions of types of evidence originating from the US Agency for Health Care Policy and Research. Furthermore, the stress distribution of a Zirconia implant to the surrounding bone could be associated with even higher stress peaks compared to Titanium, due to the higher elastic modulus of Zirconia of 210 GPa. The aforementioned pitfalls and other drawbacks of ceramics induced researchers willing to avoid metals as well to seek polymeric materials that are conveniently used in biomedical applications.

Among them, poly-ether ether- ketone (PEEK) has already been employed to replace metallic implant components in the field of orthopaedics, traumatology, and for calvarial reconstructions, where the mechanical conditions differ from those traditional bone-anchored dental implants. Presently, in the field of dentistry, clips on implant bars and healing abutments are sometimes manufactured recurring to PEEK. 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. However, PEEK alone is generally bioinert and is not conducive to cell adhesion. Recent studies have proposed new processing and

surface modifications that affect the biological and mechanical properties of pure PEEK.

CONCLUSION

In evaluating the present and predicting the future, one must also reconsider the past. The implant materials, their composition and properties are not talked about in most of the implant related literature. Modern dentistry is beginning to understand, realize, and utilize the benefits of biotechnology in health care. Study of material sciences along with the biomechanical sciences provides optimization of design and material concepts for surgical implants. For all these reasons, some new promising materials have been considered for dental implants. Ti alloys (Ti-Zr and Ti-20Nb-10Zr-5Ta) and Zirconium alloy (amorphous alloy) possess good mechanical properties, exhibit good biocompatibility, and have been proposed as an alternative to cpTi and Ti6Al4V. Similarly, Zirconia toughened Alumina (ZTA) and Alumina Toughened Zirconia (AZT) have been considered an alternative to Zirconia. They can easily processed by power injection molding (PIM), and ATZ materials show increased mechanical stability and improved aging resistance versus Y-TZP. Among polymeric materials, poly-ether-ether-ketone (PEEK) is a viable option for dental implant manufacturing, being colorless and endowed with an elastic modulus close to that of the bone. However, PEEK alone is generally bioinert and needs to be modified. Anyway, further investigations and clinical results on safety are necessary for these materials in order to consider them as possible substitutes of Titanium and Zirconia for dental implants manufacturing.

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