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Review Article

A COMPLETE REVIEW OF DENTAL IMPLANT MATERIALS

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ABSTRACT

The goal of modern dentistry is to restore the patient's normal contour, function, comfort, esthetics, speech and health. In particular the field of implant dentistry has significantly changed over the years. With the on-going science and research in the field of implantology, treatment has moved on from single tooth implant replacements to complex restorations with multiple implant positioning, enhancing implant success and improving clinical outcomes. The long-term success of dental implants largely depends on rapid healing with safe integration into the jaw bone. Geometry and surface topography are crucial for the short and long-term success of dental implants. Implant surfaces have been developed in the last decade in a concentrated effort to provide bone in a faster and improved osseointegration process. Several surface modifications have been developed and are currently used with the aim of enhancing clinical performance.

INTRODUCTION

Implant dentistry is the second oldest discipline in dentistry. Root form implants history dates back thousands of years and includes civilizations such as ancient Chinese who, 4000 years ago, carved bamboo sticks in the shape of pegs and drove them into bone for fixed tooth replacement. The Egyptian, 200 years ago, used precious metals with a similar peg design. Incas from North America took pieces of sea shells and tapped them into bone to replace missing teeth.[1-6](Fig 1)

Figure 1

Maggiolo introduced the more recent history of implant dentistry in 1809 with the use of gold in the shape of a tooth root.[7] In 1887, Harris reported the use of teeth made of porcelain into which lead-coated platinum posts were fitted.[8] Many materials were tested and in the early 1900s, Lambotte fabricated implants of aluminium, silver, brass, red copper, magnesium, gold and soft steel plated with gold and nickel.[9] The first root form design that differed significantly from the shape of a tooth root was the Greenfield lattice-cage design in 1909, made of iridoplatinum.[10] This was also the first two-piece implant, which separated the abutment from the endosteal implant body at the initial placement. In 1946, Strock designed the first titanium, two-piece screw implant that was initially inserted without the permucosal post and individual crown were added after complete healing.[11] The desired implant interface described by Strock was a direct bone implant connection, which was called ankylosis [12-17].

Implant Biocompatibility

With respect to metals, commercially pure (c.p) titanium, niobium and possibly tantalum are known to be most well accepted in bone tissue. In the case of c.p. titanium, there is likewise a documented positive long term function. The reason for the good acceptance of these metals does probably relate to the fact that they are covered with a very adherent, self-repairing oxide layer which has an excellent resistance to corrosion. Other metals such as different cobalt-chrome-molybdenum alloys and stainless steels have demonstrated less good take in the bone bed, but it is uncertain if this is valid for every possible such alloy and if it is biocompatibility effect

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alone that is responsible for their less satisfactory incorporation into bone, compared with c.p. titanium. A significantly impaired interfacial bone formation compared to c.p. titanium has been found with titanium-6 aluminium-4 vanadium alloy, probably dependent on a less good biocompatibility of the alloy. One concern with metal alloys is that one alloy component may leak out in concentrations high enough to cause local or systemic side effects. Ceramics such as the calcium phosphate hydroxyapatite (HA) and various types of aluminium oxides are proved to be biocompatible and due to insufficient documentation and very less clinical trials, they are less commonly used. With respect to HA, the available literature points to at least a short term (<10 weeks) enhanced interfacial bone formation in comparison to various reference metals. This represents a potential clinical benefit of HA, whereas the risk or coat loosening with subsequent problems represents a potential risk[18].

Implant Materials

Materials used for the fabrication of dental implants can be categorized in 2 different ways. From a fundamental chemical point of view, dental implants fall into 1 of the following 3 primary groups: metals, ceramics, and Polymers. In addition, biomaterials can be classified based on the type of *biologic response* they elicit when implanted and the long-term interaction that develops with the host tissue. Three major types of biodynamic activity have been reported: (1) biotolerant, (2) bioinert, and (3) bioactive. The different levels of biocompatibility emphasize the fact that no material is completely accepted by the biologic environment. To optimize biologic performance, artificial structures should be selected to minimize the negative biologic response while ensuring adequate function. Biotolerant materials are those that are not necessarily rejected when implanted into living tissue, but are surrounded by a fibrous layer in the form of a capsule. Bioinert materials allow close apposition of bone on their surface, leading to contact osteogenesis. Bioactive materials also allow the formation of new bone onto their surface, but ion exchange with host tissue leads to the formation of a chemical bond along the interface (bonding osteogenesis). Bioinert and bioactive materials are also called *osteoconductive,* meaning that they can act as scaffolds allowing bone growth on their surfaces. *Osteoconductive* should not be confused with osteoinductive materials, such as recombinant human bone morphogenetic protein 2 (rhBMP-2), which refers to the capacity to induce bone formation de novo. Biotolerant, bioinert, and bioactive materials are all biocompatible by definition and result in a predictable host response in specific application, *Biomimetics* are tissue engineered materials designed to mimic specific biologic processes and help optimize the healing/regenerative response of the host microenvironment. Biomimetic materials can be any combination of the chemical and biodynamic, activity categories, depending on the therapeutic strategy and the type of host tissue[19,20].

Tissue Interactions

Oxide modification during in vivo exposure has been shown to result in increased titanium oxide layer thickness of up to 200 nm. The highest oxide growth area corresponded to a bone marrow site while the lowest growth was associated with titanium in contact with cortical regions of bone. Increased levels of calcium and phosphorus were found in the oxide surface layers and seemed to indicate an active exchange of ions at the interface. Hydrogen peroxide environmental condition has been shown to interact with Ti and form a complex gel. “Ti gel conditions” are credited with attractive in vitro properties such as low apparent toxicity, inflammation, bone modeling, and bactericidal characteristic. The surface biointeraction processes may be slow or activated by local reactions and may cause ion release and oxide alteration of the substrate. Local and systemic increases of the ion concentration have been reported. In vitro studies showed that both Ti and Ti alloy were released in measurable quantities of the substrate elements at the surface. Ion release corresponds to an oxide layer thickness growth with inclusions of calcium, phosphorus, and sulfur in particular. This is especially a concern for larger orthopedic or porous implants where such ion release may be a part of the origin of implant failure, allergic reactions, and even proposed to be a local or systemic reason for the formation of tumors. In addition, free Ti ions have been shown to inhibit the growth of HA crystals (i.e., the mineralization of calcified tissues at the interface.)

Integration with Titanium and Alloys

Although titanium is known to exhibit better corrosion resistance, independent of the surface preparation, in vivo and in vitro studies have shown that titanium may interact with the recipient living tissues over several years. This interaction results in the release of small quantities corrosion products even though there is a thermodynamically stable oxide film. Lemons studied single-stage solid implants modified by bending or cutting and showed that damage could increase corrosion. Cohen and Burdairon showed that odontologic fluoride gels which create an acidic environment, can lead to the degradation of the titanium oxide layer and possibly inhibit osseointegration process. Protocols for manufacturing and cleaning prosthetic titanium parts (specifically abutments contacting the implant body) appear less stringent than those for implant bodies. This should not be the case and the same standards should be applied to both implant body and prosthetic components. In addition, the short and longer clinical implications of the potential galvanic corrosion effect could be ideally nullified by the use of electrochemically compatible alloys for the superstructure[21].

Cobalt and Iron Alloys

The alloys of cobalt (Vitallium) and iron (surgical stainless steel - 316L) exhibit oxides of chromium (primarily Cr_2O_3 with some sub oxides) under normal implant surface finishing conditions after acid or electrochemical passivation. These chromium oxides, as with titanium and alloys, result in a significant reduction in chemical activity and environmental ion transfers. Under normal conditions of acid passivation, these chromium oxides are relatively thin (nanometer dimensions) and have an amorphous atomic structure. The oxide atomic spatial arrangement can be converted to a crystalline order by elevated temperature or chemical exposures. The chromium oxides on cobalt and iron alloys are microscopically smooth, and again, roughness is usually introduced by substrate processing (grinding, blasting, or etching). Because these oxides, similar to titanium oxides, are very thin (nanometer dimensions), the reflected light color of the alloys depends on the metallic substrate under the oxide. However, as mentioned, the titanium, cobalt, and iron metallic systems depend on the surface reaction zones with oxygen (oxides) for chemical and biochemical inertness. The iron-based alloy chromium oxide and substrate are more susceptible to environmental breakdown, in comparison to cobalt and titanium-based biomaterial. The noble alloys when used in a polished condition are resistant to debris accumulation on a relative basis compared with other alloys. This has been listed as an advantage for their use in intraoral abutment systems. Also, mechanical finishing of the more noble alloys can result in a high degree of polish and a minimal concern about damaging or removing surface oxides.

Ceramics

Aluminum oxide (Al_2O_3) ceramics are fully oxide materials, bulk, and surface, thereby affording advantages related to tissue interface related investigation. Also, studies have included the polycrystalline (alumina) and single crystalline (sapphire) forms of the oxide structure. These forms have introduced very different surface roughness values for the same material substrate plus bulk properties where ion transfer and electrochemical phenomena are minimal influences. The aluminum oxides are crystalline and extend throughout the surface and bulk zones, biomechanical instabilities do not alter the chemical aspects of biomaterial properties. (No electrochemical change is introduced if the surface is removed.) Ceramic coatings (Al_2O_3) have been shown to enhance the corrosion resistance and biocompatibility of metal implants, in particular surgical stainless steel and Ni-Cr, Co-Cr alloys. However, the Ni-Cr and steel alloys can be subject to crevice corrosion.

Calcium phosphate ceramics

The calcium phosphate ceramics used in dental reconstructive surgery include wide range of implant types and thereby a wide range of clinical applications. Mixtures of particulate with collagen, and subsequently with drugs and active organic compounds such as bone morphogenic protein increased the range of applications. The coatings of metallic surfaces using flame or plasma spraying (or other techniques) increased rapidly for the $CaPO_4$ ceramics. The coatings have been applied to a wide range of endosteal and subperiosteal dental implant designs with an overall intent of improving implant surface biocompatibility profiles and implant longevities. The advantages of $CaPO_4$ ceramic biomaterials are: Chemical compositions of high purity and of substances that are similar to constituent of normal biological tissue. Excellent biocompatibility profiles with in a variety of tissues when used as intended. Minimal thermal and electrical conductivity plus capabilities to provide a physical and chemical barrier to ion transport. Moduli of elasticity similar to bone than many other implant materials used for load bearing implant.

$CaPo_4$ ceramic properties

The crystalline monolytic hydroxylapatite (Fired ceramic HA) of high density and purity has provided one standard for comparison and purity has provided one standard for comparison related to implant applications. Considerble differences exist between the synthetic HA ceramics (Hydroxylapatite) that are produced by elevated temperature processing and biological apatites (Hydroxyapatites). Biological apatites contain trace amounts of CO_3, sodium, Mg, F and Cl ions. The crystalline Tricalcium phosphate ceramic has also provided a high purity biomaterial for comparison with other products. The HA particles can have relatively high compressive strengths, with tensile strengths in the range of 50-70 Mpa. The coatings of $CaPo_4$ ceramics onto metallic (cobalt and Ti-based) biomaterials have become a routine applications for dental implants. The coatings of thicknesses between 20 and 100 µm; are mixtures of crystalline and amorphous phases and have variable microstructures compared with the solid portions of the particulate forms of HA and TCP biomaterials. The $CaPo_4$ coatings are nonconductors of heat and electricity. Relative solubilities of $CaPo_4$ ceramics have been determined and is greater for TCP than for HA. Solubility depend on environment (like pH, mechanical motion and so forth) . In most application solubilities are higher over the first few weeks; then decrease with continued in vivo exposure and the apposition of mineralized structures. However some investigators have shown situations where osteoclastic resorption has removed localized zones of $Ca\text{-}PO_4$ coatings.

Polymers and composites

The utilization of synthetic polymers and composites continues to expand for biomaterial applications. Fiber reinforced polymers offer advantages in that they can be designed to match tissue properties, can be anisotropic with respect to mechanical characteristics, can be coated for tissue attachment, and can be fabricated at relatively low cost.

Structural biomedical polymers

The more inert polymeric biomaterials include polytetrafluoroethylene (PTFE), Polyethylene pterephthalate (PET), Polymethylmethacrylate (PMMA). Ultrahigh molecular weight polyethylene (UH<W-PE) Polypropylene (P.P.) Polysulfone (PSF) and polydimethylsiloxane (PDS or silicone rubber). In general the polymers have lower strengths and elastic moduli and higher elongations to fracture compared with other classes of biomaterials. They are thermal and electrical insulators and are relatively resistant to biodegradation. Polymers have been fabricated in porous and solid forms for tissue attachment, replacement and augmentation and as coatings for force transfer to soft tissue and hard tissue regions. Cold flow characteristics and creep and fatigue strengths are relatively low for some classes of polymers. (Eg. PMMA and some are extremely tough and fatigue cycle resistant (PP, UHMW-PE, PTFE) and afford opportunities for mechanical force transfer in selected implant designs[9].

Zirconia

Zirconia (ZrO_2) is a ceramic material used in implantology because of its biocompatibility, esthetics, and mechanical properties, which are better than alumina. Implants made of zirconia are inert, radioopaque, and present a high resistance to corrosion, flexion, and fracture. It presents a contact with bone and tissue similar to that observed in titanium implants, and it can be used to produce an entire implant or as a surface coating. The interface is composed of proteoglycan layer, which is thicker than titanium (ranging from 300 to 500 A° and

200 to 400 A°). However, the amount of bone formed 1 and 6 months after implant placement (in rabbits) did not differ in titanium and zirconia implants. Bone response to ZrO2 implants was evaluated in a rabbit study, Four weeks after implantation, the BIC value was 68.4% and the authors reposted an absence of epithelial down growth, foreign body reaction, gaps, or fibrous tissue between bone and implant. The stability of osseointegration around ZrO2 implants was also evaluated under different loading conditions in a monkey model. Dental implants were inserted, and 3 months later, prosthesis were installed (single freestanding implant support, and a combination of implant and tooth support). Peri-implant tissues were observed by clinical, histologic and histomorphometric examination 12 and 24 months after loading, and statistically significant differences were observed among groups (BIC values ranged from 66% to 81%) [9].

Physical and chemical treatments of zirconia were shown to largely influence its soft tissue interactions (mainly fibroblastic ones). Moreover, few studies highlighted that zirconia and its derivatives (ZrN) have the capacity to reduce plaque on implant and surrounding tissues and consequently should be important in soft tissue healing and implant success at bone level. It probably avoids the resorption of peri-implant bone as well. Finally, the capacity of zirconia to be colored to match natural teeth tint appeared to be a beneficial property compared to titanium in aesthetical demanding regions. The future of dental implantology should aim at developing a serious modification of production zirconia processes to get surfaces with controlled and standardized topography or chemistry. This approach will be the only way to understand the interactions between proteins, cells and tissues, as well as implant surfaces. This strategy should ultimately enhance the osseointegration process of dental implants for their immediate loading and long-term success. Finally, new zirconia-based composite bioceramics are under investigation, that is, hydroxyapatite-zirconia or titania-Y-TZP[23-26].

CONCLUSION

The endosseous dental implant has become a scientifically accepted and well documented treatment for fully and partially edentulous patients. Titanium and its alloys are the materials of choice clinically, because of their excellent biocompatibility and superior mechanical properties. Endosseous tapered and screw shaped dental implants are currently preferred due to their threads engaging in the bony walls which allows for good primary stability and the threads increasing the surface area in contact with bone. Thread pitch should be minimal (increased amount of threads) in order for best resistance to vertical loading With regard to microscopic features, titanium is considered the material of choice due to its inert processes and it does not inhibit osteoblast growth. Titanium alloys are used to improve the strength characteristics. For surface morphology, a roughened surface results in an increased BIC and a decrease in the shear forces observed. The composite effect of surface energy, composition, roughness, and topography on implant determines its ultimate ability to integrate into the surrounding tissue.

Future development of the next, third generation of dental implants should be based on increased knowledge about the interface biology on cellular and molecular levels. The development of future generations of oral implants for compromised tissue conditions will, most probably, entail tailored modifications of material surfaces. Implant surfaces, selectively, designed for drug and/or cell releases represent promising candidate strategy. Other surface modifications, such as selective ion substitutions of biomimetic surfaces may further improve the biological response to those surfaces. Further, as the bacterial infection is a major challenge which may jeopardize the success of osseointegrated implants, implant modification resulting in antibacterial activity might be of importance to reduce such complications. sThus, the continuing search for ‘‘osseoattractive’’ implants is leading to several surface modifications which will promote interaction of mucosal and submucosal tissues with dental implants.

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