# Monitoring, Control and Prevention Practices of Biomaterials Corrosion – An Overview

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*Biomaterials are materials used for many devices that can interact with biological systems to coexist for longer service with minimal failure. This paper provides an account of the chemical stability and biocompatibility in body environment of biomaterials. Corrosion is one of the major processes affecting the life and service of biomaterials devices made of metals and alloys. It has been observed that biomaterials corrosion occurs mainly by electrochemical reactions. This paper highlights the major corrosion processes. In order to reduce corrosion and achieve better biocompatibility; design, materials selection, and surface modifications are considered potential methods. The electrochemical techniques of mitigating corrosion are hereby examined in depth. © Society for Biomaterials and Artificial Organs (India), 20090113-36.*

## Introduction

The intersection of biomedical science and materials engineering is an exciting one and falls predominantly in the province of biomaterials and tissue engineering. Biomaterials are materials that are used in medical devices or are in contact with biological systems and do not adversely affect the living organism and its components (Forysthe, 2007). Delong (2005) stated that the biomaterials area is focused on synthesis, structure, and properties of novel materials and nanostructures. Specifically, it will address either the mimicking of natural materials, using organism as biomaterials factories of new materials, genetically altering existing organisms for new materials capabilities, or taking existing biomaterials or organisms, and using them as novel material like viral ingredients or processing them further to make useful material as in biomineralization.

Biomaterials are either modified natural or synthetic materials, which find application in a wide spectrum of medical and dental implants and prosthesis for repair, augmentation or replacement of natural tissues. Some well-known examples of the clinical use of biomaterials are total joint replacement, vascular grafts, heart valves, cranial plates, Harrington rods, artificial tendon and ligament, bone space fillers and disc crest repair (Bonfield, 2004).

*Socio-economic Consequences:* The field of biomaterials is rapidly expanding. It is considered that this domain represents 2-3% of the overall health expenses in developed countries. For example, France spends every year 30-40 billions of French francs. World expenses could be estimated at 100 billions of dollars (Sedel, 2004). In Australia, it is thought that biomaterials-based devices cost some $AUS 400 billion dollars per year which constitute roughly 8% of money spent on health related issues. Even a specialist business such as making spinal implants is currently a $2.5

billion industry today but is expected to involve some 425 billion in sales for this orthopeadic device alone in future years (Forysthe, 2007).

The number of patients requiring and receiving biomaterials to correct skeletal defects and heal diseases are constantly increasing. (Heimann, 2002). Bonfield (2004) revealed that annually in the US there are 250,000 age-related hip fractures with an estimated health care cost approaching $10 billion and 700,000 age-related vertebral fractures, which cost £1 billion. While in Britain, the equivalent overall costs approach £2 billion. The wound repair market in the US, for example, is currently $520 million, heading up to $900 million in a matter of years. One of the most widely used biomaterials-based prosthesis is the hip replacement and it is estimated that about 1,000,000 of these are implanted per year worldwide. The use of materials in biomaterials applications gives them enormous value compared to being used in other applications. For example, it's been estimated that $600,000 worth of materials used traditionally in non-bio areas, has the value of some $10.5 billion when constructed into biomedical devices (Forysthe, 2007). It must be noted that the socio-economic cost in developing countries are not documented, hence it is advocated that effort should be made in finding the consequences of biomaterials in these regions.

There are several definitions of medical device, biomaterial, surgical material and so on (William, 1999). The concept and definition of biomaterials will be extensively explained in the next section.

**Definition and Concept of Biomaterials**

Biomaterials are made from conventional materials such as ceramics, polymers, metals and alloys, and composites. They are exposed to the biochemical and dynamic environment of the human body and their design is dictated by anatomy and restricted by physiological conditions. The clinical objective of biomaterials is to relieve pain and increase ease of movement in the joint (Kamachi Mudali et al, 2003). While the engineering objective is to provide minimal physiological stress to the remaining bone system so that the integrity and

functionality of the bone and prosthetic materials are maintained over a long service life. Thus materials suitable for implantation are those that are well tolerated by the body and can withstand cyclic loading in the aggressive environment (Kamachi Mudali et al, 2003). Biomaterials are defined as a non-variable material used in a medical device intended to interact with biological systems (Ratner et al, 1996; Williams, 1981). The fundamental requirement of a biomaterial is that the material and the tissue environment of the body should coexist without having any undesirable or inappropriate effect on each other; this is known as biocompatibility

However, this is not an individual property per se but relates to the various interactions on the cell and tissue levels the material is subjected to. Hence a systems approach is required and it involve the host material and the interface (Kasemo & Lausmaa, 1986; Williams, 1985; Willliams, 1990). It is worth bearing in mind the currently accepted definition of biocompatibility which is 'the ability of a material to perform with an appropriate host response in a given situation' (William, 1999). This definition emphasizes the positive nature of the interactions. It allows for the fact that the most appropriate situation may be inertness and non-recognition, but also implies that interactivity between the material and the host could be positively encouraged and directed in a way that is most beneficial for the functionality and retention of the device (Williams, 2003). In increasing order of biocompatibility, the interaction of biomaterials with living tissue can be ranked as follows (Wintermantel and Ha, 1996): incompatible materials, biocompatible materials these are otherwise known as biotolerant, bioinert materials and bioactive materials.

As noted before, most biomaterials are selected on the basis of their inertness, which in this context implies a resistance to degradation. This again is a very complex matter since there are so many active substances and components of the physiological environment that the potential to interact with and ultimately degrade material surfaces (Williams, 2003).

**Biomaterials Corrosion**

Corrosion is one of the major processes that cause problems when metals and alloys are used as biomaterials in the body (Kruger, 1979). It is realized from this definition that although biomaterials deteriorate over time, metals and alloys are the only ones highly susceptible while polymers also suffers degradation. Corrosion of biomaterials is an important aspect of biocompatibility and only the noblest metals (gold and platinum group metals) or the most passive (titanium or chromium) metals have corrosion rates within apparently acceptable levels. The metallic biomaterials behaviour in bioliquids is a function of many parameters, related to surface preparation and environment specific composition including the special influence of chlorine or fluorine anion or the effect of organic compounds [ Demetrescu & Popescu, 2003; Yu & Zhao, 1993; Reclaru & Meyer, 1998; Born et al, 1998].

Corrosion of implants in the aqueous medium of body fluids takes place via electrochemical reactions (Shreir, 1994) and it is necessary to appreciate and understand the electrochemical principles that are most relevant to the corrosion processes. The electrochemical reactions that occur on the surface of the surgically implanted alloys are identical to those observed during exposure to seawater (namely, aerated sodium chloride). The metallic components of the alloy are oxidized to their ionic forms and the dissolved oxygen is reduced to hydroxyl ions. The types of corrosion that are pertinent to the currently used alloys are pitting, crevice, galvanic, intergranular, stress corrosion cracking, corrosion fatigue and fretting corrosion (Kamachi Mudali et al, 2003).

*Pitting Corrosion:* this is a severe form of localized corrosion attack, which results in extensive damage and release of significant amounts of metal ions. Pitting refers to the formation of small cavities or holes at the surface of a material, which is protected otherwise by the presence of an adherent, tenacious and self healing thin passive film. The formation mechanism and principles of pitting corrosion has in Figure 2 and these has been extensively studied by many workers [Fontana and Greene, 1987; Jones, 2000; Trethewey and Chamberlain, 1995; Colangelo & Heiser, 1974].

In implants, pitting occurs most often on the underside of screw heads. This form of attack occurs more frequently in media containing chloride ion (Sivakumar et al, 1994). It is well established that the resistance to pitting in saline environment can be increased by molybdenium addition and keeping the inclusion contents to a minimum level.

*Crevice Corrosion:* is a form of corrosion that occurs when a metal surface is partially shielded from the environment. It is usually encountered beneath the screw head that holds the plate or in similar location such as the intersection of bone plate. The requirement for the occurrence of this process is the presence of a device, such as between plate and screw head, or defects such as fatigue crack. Bates (1973) stated that type 316L is highly susceptible to crevice corrosion attack as compared to other commonly used metallic implant material. Crevice corrosion problem can often be eliminated by appropriate design of device and proper choice of the material.

*Galvanic Corrosion:* is a bimetallic corrosion that takes place when two different metals are in physical contact in an ionic conducting fluid medium such as serum or interstitial fluid. The differential composition or process variables of a plate and the adjoining screws is responsible for the set-up of a galvanic couple, which results in galvanic corrosion. It depends on a large number of complicating factors including the relative areas of electronic and ionic contact, as well as the actual metal pair involved. In many practical applications, the contact of dissimilar materials is unavoidable. In biomaterials, galvanic corrosion can occur if bone plate and bone screws are made of different metals or alloys (Kamachi Mudali et al, 2003).

*Corrosion Fatigue:* is a fracture failure of metal that occurs because of the combined interaction of electrochemical reactions and cyclic loading. Corrosion fatigue resistance is an important factor for consideration for load-bearing implant metals or for metals used in cyclic-motion applications. Normally, a failure may not occur, but cracks can initiate from hidden imperfections, surface damage, minute flaws, chemical attack and other causes. The corrosive environment may result in local corrosive attack that accentuates the effect of the various

imperfections. Failures of mechanical origin in implants are most commonly due to fatigue or environmentally assisted fatigue. Ha et al (1993) reported that the mechanism responsible for crack initiation and propagation may be different.

*Fretting Corrosion:* occurs when two opposing surfaces such as bone plates and the screw heads of the prosthetic devices rub each other continuously in an oscillating fashion in the body environment. It is the result of small relative movements between the contacting surfaces in a corrosive medium. Even in the absence of corrosive medium, fretting can occur. Clinical significance of fretting attack lies in its intensity that may give rise to a large amount of corrosion products in adjacent tissues or it may be a major factor in crack initiation and fracture failure of an implant (Syrett & Wing, 1978).

Quantification of weight loss of implant due to fretting corrosion was found to be directly proportional to the load transmitted across the surfaces, the number of cycle fretted and amplitude of stresses. The weight loss has been reported to be inversely proportional to the hardness of the material and the frequency of stroke (Kamachi Mudali et al, 2003).

Williams (2003) observed that material degradation can have two broad consequences for medical devices. First, degradation of a material can result in loss of structural integrity of a device, possibly with its ultimate dissolution or removal. This may be undesirable in the case of a device designed to be inert, but could be desirable in those devices which are intentionally biodegradable. Secondly, the release of products of the degradation process may affect the tissues, either locally or systemically, and either adversely, when unintended, or possibly beneficially when the released products have desirable and intended biological functions. The release of components from biomaterials may take different forms (Kamachi Mudali et al, 2003). In particular, the key feature of the tissue in the response to a degrading material is inflammation (Hunt et al, 1995). The presence of materials and the corrosion products tend to aggravate the inflammation and these inflamed cells are more aggressive to materials than the normal extracellular fluid, the process is likely to

become autocatalytic (Kamachi Mudali et al, 2003).

Generally there is a need for assessment of biomaterials degradation due to these adverse effects. The quantification techniques that are commonly employed are discussed in the next section but these techniques are not limited to the ones discussed.

**Monitoring Biomaterials Corrosion**

The most important requirement for an implantable material is that it must be corrosion resistant to physiological environments. All metallic implantable alloys are susceptible to corrosion to a certain degree, depending on the metallurgical condition, residual or service stresses, thermal history, and final surface treatment prior to application.

It will be recalled that biomaterials corrosion is regarded to occur via electrochemical reactions based on the principle that most monitoring techniques of the degradation process utilizes electrochemical approach. The electrochemical studies were always carried out in simulated body fluids, namely, aerated seawater, Ringer's solution, Hank solution and Phosphate Buffered Solution (PBS). The pH of the solution is always maintained at 7.4 and a temperature of 37 $\pm$ 1°C. Also some of the evaluation is carried out in-vivo or in-vitro.

The techniques used are potentiostatic polarization, potentiodynamic polarization, and electrochemical impedance spectroscopy among others while the ions count are measured by using equipment like AAS and AES. The surface characterizations are carried out by atomic force microscopy or mass spectroscopy. It must be noted that standards presuppose measurement hence in this study emphasis will be placed on ASTM Standard and about three other techniques reported will be discussed.

*ASTM Standard:* ASTM G31 is an immersion test procedure that typically requires extended exposure periods, up to years, to evaluate the resistance of materials, which are employed in the implant device industry, to corrosion. As such accelerating the testing by inducing changes on the material and monitoring the results has

been an industry-accepted method. During the selection of a material for use as an implantable device, certain tests, such as F746, G5 and G61, were used to identify corrosion resistance. All metallic implantable alloys are susceptible to corrosion to a certain degree, depending on the metallurgical condition, residual or service stresses, thermal history, and final surface treatment applied prior to implantation. It is realized that corrosion of implant devices would have detrimental effects on their performance therefore it is necessary to determine their susceptibility to localized attack (Corbett, 2005).

Thus, the ASTM committee FO4 on Medical and Surgical Materials and Devices adopted F2129 as a modification of the ASTM Standard G61. This standard assessed the corrosion resistance of small metallic devices or components using the G-61 technique but specific to implants.

Gilbert (1998a) evaluated three techniques that have been developed and used to provide detailed quantitative information concerning the electrochemical behaviours of metallic biomaterials. The three techniques are scanning electrochemical microscopy (SECM), high – speed electrochemical scratch testing, and step-polarization impedance spectroscopy. The techniques provide unique information relating to the local electrochemical and mechanical phenomena and the measurement of electrical properties of surface oxides.

*Scanning Electrochemical Microscopy (SECM):* is a technique that has evolved from other scanning probe technologies [Gilbert et al, 1993; Bard et al, 1991; Pierce & Bard, 1993]. The sample and the probe can be polarized to whatever potentials are desired in order to react with the solution or environment.

SECM has many potential applications to biomaterials research. These include the study of corrosion heterogeneity, different microstructures, the monitoring of electroactive species distribution over a surface and even the potential-time biologically based electrochemical reactions near implant surfaces or near viable cell membranes. With the scanning capabilities of SECM, one may be able to spatially and temporally resolve electrochemical events at or near cells while

they are situated on a material surface. It is believed that this technique is promising and may provide insights into corrosion behaviour or other electrochemical processes relevant to biomaterials research.

*Electrochemical Scratch Tests:* most alloys used in medicine and dentistry rely on a surface oxide film towards protection against corrosion. In the absence of mechanical factors, these oxide films are in fact highly corrosion resistance. However, when stress, fretting or wear are present, the corrosion can increase as well (Gilbert et al, 1993; Gilbert & Jacobs, 1997). Electrochemical scratch test is the most suited for evaluation of the combined role of mechanical and electrochemical factors. When the area scratched $(A_0)$ is known, then the thickness (ä) can be determined as shown below:

$$ä = (q_{film}M_w) / ñZFA_0$$

where, ñ is the density of the oxide film, z is the charge per cation, F is Faraday's constant, and $M_w$ is the molecular weight of the oxide.

A significant advantage of the scratch test system is that it is capable of directly imaging alloy surface topography before and after a scratch and so can directly measure the depth of a scratch resulting from a set of conditions (e.g. applied load, scratch length, solution, potential etc). The technique is thus ideal for evaluating new surface treatments or alloy variables and for investigating the role of varying such chemistries on the behaviour of alloy surfaces exposed to both mechanical wear and electrochemical corrosion.

*Step-Polarization Impedance Spectroscopy (SPIS):* In this technique the electrical nature of the alloy-solution interface is of primary importance. Most classical electrochemical investigations involve either polarization testing, in which the current-potential relationship for an alloy is monitored, or testing of impedance using electrochemical impedance spectroscopy methods.

SPIS is a technique whereby both polarization and impedance behaviour can be simultaneously determined over a wide range of potentials. The polarization response is

simply the potential versus the current density taken at any time after the step in potential is applied. The impedance characteristics on the other hand are derived from the current transient after it is subjected to a piece-wise numerical laplace transition technique, which converts the time-domain data to the frequency domain (Gilbert, 1998b). These results in a quantity called the admittance ($A^*_{[ù]}$), a complex number containing both real (A') and an imaginary (A") part:

$$(A^*_{[ù]}) = 1/Z^*_{(ù)} = A'_{(ù)} + iA"_{(ù)}$$

The admittance data provide detailed information regarding the electrical nature of the interface which is used in biomaterials to assess the behaviour of oxide films as well.

There is a need to put in place measures that will mitigate corrosion of biomaterials in bioliquids as such some of the most employed methods are presented in the following section.

**Control and Prevention of Biomaterials**

The preventive measures for corrosion system include: design, materials selection, surface modification (inhibitions and coatings), materials modification (by alloying and or heat treatment) and electrochemical techniques (cathodic protection and anodic protection). In biomaterials corrosion, all the methods are applicable with the exception of electrochemical techniques which have not been extensively recorded.

*Design:* Design engineers must consider the physiological loads to be placed on the implants, so they can design for sufficient structural integrity. Material choices also must take into cognizance the biocompatibility with surrounding tissues, the environment and corrosion issues, friction and wear of the articulating surfaces, and implant fixation either through osteointegration (the degree to which bone will grow next to or integrate into the implant) or bone cement.

It is observed that most of the corrosion attack that biomaterials are susceptible to, design plays a major role in their prevention. For example, galvanic corrosion are prevented by avoiding two dissimilar metals while pitting

and crevice corrosion place emphasis on little or no shielded regions. Pits, cavities, inclusions, among others are also to be avoided (Jones, 1992; Trethewey & Chamberlain, 1987).

*Materials Selection:* This is the most common methods of combating biomaterials corrosion. The widely used biomaterials are classified into three major classes: metals, polymers and ceramics. Recently, composites are also being designed for certain biomedical applications (Quellete, 2001).

*Metals:* The fundamental criterion for choosing a metallic implant material is that it should possess biocompatibility. Metals and alloys have been widely used in various forms as implants, which provide the required mechanical strength and reasonable corrosion resistance. Metallic implants are usually made of one of the three types of materials: austenitic stainless steels, cobalt-chromium alloys, and titanium and its alloys (Sivakumar et al, 1992; Sivakumar et al, 1994). Amongst all these materials, titanium and its alloys are the most corrosion resistant materials.

*Titanium:* is as strong as steel, but much lighter and it is important to check the composition of the first layers of the titanium implants. This will determine the assessments of the electrochemical reactions at the surface and the interface to living tissue. In the living, titanium has a very high polarization resistance. Ti-6Al-4V, Ti-5Al-2.5Fe, and Ti-6Al-7Nb are the titanium alloys used in biomaterials. The main disadvantages are their high cost, inferior wear properties, diffusion of oxygen into titanium during fabrication and the effect of heat treatment and the dissolved oxygen that embrittle titanium. Special fabrication and welding procedures, heat treatment and surface modification are used to enhance their properties (Vidal Otero et al, 2005; Kamachi Mudali et al, 2003; Putter et al, 1969).

*Stainless Steel:* this is the most common biomaterials that had been used for many years, especially type 316L. Stainless steels are characterized by their strength, hardness, corrosion resistance, biocompatibility, fatigue resistance, and ease of sterilization. There uses are however limited because they are prone to crevice and fretting corrosion(Vidal

Otero et al, 2005; Kamachi Mudali et al, 2003; Putter et al, 1969).

*Nitinol:* nitinol is the most commercially successful shape memory alloys (SMAs) and they are important materials for biomedical and dental devices because of their unique properties, comparatively high corrosion resistance and good biocompatibility.

The good corrosion resistance of nitinol results from the formation of very stable, continuous, highly adherent, and protective oxide films on its surface. Because titanium is highly reactive and has a high affinity for oxygen, these beneficial surface oxide films form spontaneously when fresh metal surfaces are exposed to air and/or moisture. In fact, a damaged oxide film can generally reheal itself if at least traces of oxygen or water (moisture) are present in the environment (Stone et al, 1999). In addition, calcium phosphate surface films can be naturally formed on titanium alloys in a biological environment [Hanawa & Ota, 1991; Hanawa, 1999; Demri et al, 1997] which can act as a further barrier against ion diffusion from the subsurface alloy.

It is necessary to note that due to high nickel percentage, nitinol is capable of eliciting toxic and allergic responses. The compatibility and corrosion susceptibility can be optimized by heat treatment and ion implantation parameters [Tan et al, 2003; Green et al, 1993; Wu et al, 1997; Fu et al, 2000; Chern et al,1995; Riviere, 1998; Mandl et al, 2001; Tan & Crone, 2002].

The mechanical properties of the implant alloys and human bone are as presented in table 2. Generally, the use of metal is on the decline. Tissues are cellular, heterogeneous, anisotropic, aqueous and viable. Metals are acellular, relatively homogeneous and isotropic, anhydrous and hydrophobic and dead. The emerging approaches are using natural materials as biomaterials and also using structures and techniques that facilitate regeneration (Williams, 2003).

*Polymers:* are considered in various forms such as fibres, textiles, rods and viscous liquids in biomedical applications. They undergo degradation -molecular and oxidative- in the body environment due to biomechanical and mechanical factors. This results in ionic attack and formation of hydroxyl ions and dissolved oxygen leading to tissue irritation and decrease in mechanical properties. Hence there is a need for polymers that were inert as possible and substantially free of additives and contaminants.

Polytetrafluoroethylene (PTFE) is considered the most inert of all polymers and it is known to have excellent biocompatibility under many different circumstances. It does not have very good mechanical properties and this leads to development of expanded PTFE, widely known commercially as GoreTex (Williams, 2003). Silicones are used as adhesives, coatings or encapsulants in medical devices. The most important types are fluids, gels and elastomers (rubbers) and they are highly cross-linked polysiloxane networks swollen with polydimethylsiloxane (PDMS) (Vidal Otero et al, 2005; Putter et al, 1969).

The development of injectable mixture of polymers, human cells and growth stimulators that solidify and form healthy tissue is considered as a potential milestone in biomaterials development and applications (MIT, 2003). This development has a great promise as a replacement for the common, invasive and expensive surgical procedure for replacing the human knee and hip joints with artificial implants (Afonja, 2006). Biodegradable polymer is another subdivision of the polymer group frequently used in biomedicine. This usually degrades when placed in the body while allowing functional tissue to grow in its place [Oyatogun & Adeoye, 2006; Voldman, 2003; Quellette, 2001]. It must be recognized that theses group of injectable polymers can be likened to inhibitors.

Lastly the search for better and more appropriate polymers for implantation has not been straightforward and a variety of scientific and logistic difficulties has restricted these developments (Williams, 2003).

*Ceramics*: Bioceramics have become a diverse class of biomaterials presently including three basic types, viz; bioinert high strength ceramics (alumina, zirconia and carbon); bioactive ceramics which form direct chemical bonds with bone or even with soft tissue of a living

organisms (bioglass and glass ceramics); and various bioresorbable ceramics that actively participate in the metabolic processes of an organism with the predictable results (calcium phosphate ceramics) [Dubok, 2000; Thamaraiselvi & Rajeswari, 2004]. Table 3 elicits the biomedical applications of Bioceramics.

*Alumina (Al$_2$O$_3$):* an alumina ceramic has characteristics of high hardness and high abrasion resistance. The reasons for the excellent wear and friction behaviour of alumina are associated with the surface energy and surface smoothness of this ceramic. Abrasion resistance, strength and chemical inertness of alumina have made it to be recognized as a ceramic for dental and bone implants. The biocompatibility of alumina ceramics has been tested by many researchers [Noiri et al, 2002; Sawada & Ika, 1998; Kondoh & Zairyo, 1990; Kikuta et al, 1992; Yuhta et al, 1994; Kanematsu et al, 1985]. Alumina ceramic induces weak tissue reaction (Sawada & Ika, 1998).

*Zirconia (ZrO$_2$):* zirconia is a biomaterial that has a bright future because of its high mechanical strength and fracture toughness. Zirconia ceramics have several advantages over other ceramic materials due to the transformation toughening mechanisms operating in their microstructure that       can be manifested in components made out of them. They are used in total hip replacement (THR).

*Carbon:* carbon is a versatile element and exists in a variety of forms. Bokras et al (1992) emphasized that the good compatibility of carbonaceous materials with bone and other tissue and the similarity of the mechanical properties of carbon to those of bone indicate that carbon is an exciting candidate for orthopedic implants. Unlike metals, polymers and other ceramics, these carbonaceous materials do not suffer from fatigue. However, their intrinsic brittleness and low tensile strength limits their use in major load bearing applications. It is used as biomaterial particularly in contact with blood. Hence, it is important to evaluate its blood compatibility (Thamaraiselvi & Rajeswari, 2004).

*Bioglass and Glass Ceramic:* bioglasses are interesting versatile class of materials and

structurally all silica-based glasses have the same basic building block (silicate ion). Glasses of various compositions can be obtained and they show very different properties. Bioglasses have also found a place in prosthesis. These bioglasses are embedded in a biomaterial support to form prosthetics for hard tissues. Such prosthetics are not only biocompatible; they show excellent mechanical properties and are useful for orthopedic and dental prosthetics (Ire, 1995). The biocompatibility of bioglass powders were deduced from studies carried out both in vivo and in vitro.

The glass ceramic has superior mechanical properties, good biocompatibility, bioactivity and no toxicity making it useful as a biomaterial in artificial bone and dental implants (Suh & Kim, 1997).

*Ceramic/Ceramic Composite:* composites made of Bioinert and bioactive ceramics are produced to achieve two important features, bioactivity and mechanical strength. Such composites were biologically evaluated by workers through several animal tests. Alumina ceramic can form composites with hydroxyapatite (HAP) that are bioactive. Animal experiments of HAP/Al$_2$O$_3$ composite reveal that it can form tight osteointegration with bone. It is bioactive with high strength (Zeng et al, 1993).

*Calcium Phosphate Ceramics (CPC):* the most widely used calcium phosphate based bioceramics are hydroxyapatite (HAP) and â-tricalcium phosphate (â-TCP). HAP has chemical formula Ca$_{10}$(PO$_4$)$_6$(OH)$_2$, the Ca/P ratio being 1.67 and possesses a hexagonal structure. It is the most stable phase of various calcium phosphates. It is stable in body fluid and in dry or moist air up to 1200°C and does not decompose and has shown to be bioactive due to its resorbable behaviour. For biomedical purposes, the carbonated apatite and fluoapatite are the materials of interest because of assumed similarity to bony apatite and decreased solubility in aqueous solutions respectively.

â-tricalcium phosphate (â-TCP) is represented by the chemical formula Ca$_3$(PO$_4$)$_2$, the Ca/P ratio being 1.5. The main differences with HAP are carbonate and hydrogen phosphate

substitution for phosphate and the important amount of cationic and anionic vacancies. The Ca/P ratio is constant.

The most important properties of calcium phosphate biomaterials are their bioresorption and bioactivity. These phenomena are essentially dynamic and strongly depend on biological parameters (Driessens, 1988; Thamaraiselvi & Rajeswari, 2004).

*Composites:* a typical example is the combination of bioinert and bioactive ceramic materials but it is essential that each component of the composite must be compatible so as to avoid degradation between interfaces of the constituents. Fibre-reinforced polymers (FRP) are the most widely investigated composites for biomedical applications. They are characterized by uncertain lifetimes and degradation under complex states of stress and their low mechanical strengths limit their applications. It is also difficult to shape them, and they are yet to attain technical maturity for these applications (Kamachi Mudali et al, 2003).

*Surface Modification:* surface modification of biomaterials is another alternative for enhancing their corrosion resistance, biocompatibility and mechanical properties. Surface modification techniques include the following: ion implantation or deposition of ceramic layers (TiN, DLC, $Al_2O_3$, $ZrO_2$), by plasma spraying, chemical vapour deposition (CVD), physical vapour deposition (PVD) etc (de Groot et al, 1998). Surface modification is the major current areas of research in biomaterials because of their versatility but they are limited for implants due to their inferior abrasion and wear properties.

*Ion Implantation:* ion implantation is a process which involves the introduction of a small and economical amount of atoms of any element into the surface of the material with a beam of high velocity ions, without modifying the surface finish or the bulk properties of the underlying material and is independent of thermodynamic constraints. Ion implantation is a versatile surface alloying techniques which produce novel meta-stable solid solution surface alloys without any compositional limitations normally imposed by equilibrium phase diagrams (Thamaraiselvi & Rajeswari, 2004). This

surface modification enhances significantly new bone formation under in vivo condition (Ukai & Murakami, 2001).

*Coatings:* there are several technologies available for coating a material. In biomaterials, coatings are used because they impart satisfactory corrosion resistance to materials with intrinsically inadequate properties. It must be noted that several techniques of applying coats exist so also coating materials. Some of the coatings are listed as follow: hydroxyapatite coating (Lin et al, 2001; Chu et al, 2001; pfaff et al, 1993), bioceramics (Hench, 1991; Lugscheider et al, 2001), sol-gel coatings (Chai et al, 2001), quasicrystal thin films (Symko et al, 2001), functionally graded materials coatings (Khor & Wang, 2001), graded coatings (Park & Condrate, 1999), composite coatings (Wang et al, 2000; Breme et al, 2000) and biomimetic coatings (Li et al, 2001; Li et al, 2002).

## Conclusion

This review of biomaterials and its corrosion has demonstrated the significant progress and limitations that has been made with the use of advanced materials within the human body. Biomaterials corrosion is restricted to metals and alloys, and it is regarded to take place via electrochemical reactions. An attempt has been made to highlight the main corrosion processes associated with biomaterials, and the ways by which they can be mitigated. The current research areas focused on surface modification of alloys by novel methods and processes which enhance the corrosion resistance and biocompatibility of these biomedical devices. Also polymeric materials are used as injectable which can be degradable or otherwise but are highly biocompatible.

It is recommended that effort should be made to utilize electrochemical control and prevention procedures in biomaterials corrosion.

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