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Biomedical Implants: Corrosion and its Prevention - A Review

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Abstract: In the area of materials science, corrosion of biomaterials is of paramount importance as biomaterials are required for the survival of the human beings suffering from acute heart diseases, arthritis, osteoporosis and other joint complications. The present article discusses various issues associated with biological corrosion of different kinds of implants used as cardio stents, orthopedic and dental implants. As the materials used for these implants are manifold starting from metallic materials such as stainless steel (SS), cobalt chromium, titanium and its alloys, bioceramics, composites and polymers are in constant contact with the aggressive body fluid, they often fail and finally fracture due to corrosion. The corrosion behavior of various implants and the role of the surface oxide film and the corrosion products on the failure of implants are discussed. Surface modification of implants, which is considered to be the best solution to combat corrosion and to enhance the life span of the implants and longevity of the human beings is dealt in detail and the recent advances in the coating techniques which make use of the superior properties of nanomaterials that lead to better mechanical properties and improved biocompatibility are also presented.

Keywords: Biomaterials, biocorrosion, stainless steel, cobalt chromium, titanium, titanium alloy, magnesium, composites, polymer, failures, oxide layer, surface modification, cardio vascular, dental and orthopedics.

INTRODUCTION

The field of biomaterials is of immense importance for the mankind as the very existence and longevity of some of the less fortunate human beings, who even at the time of birth are born with congenital heart disease and also for the aged population who require biomedical implants to increase their life span. The aged people need the help of geriatric physicians for several ailments as the parts of the human system have performed their expected tasks for long years and have become worn out. Arthritis is one of the major illnesses generally faced by the aged and even at times young people are also affected by this disease and it impairs the life of those affected leading to immobility and unbearable pain. However, the cause of this disease remains unknown even today in spite of tremendous scientific advancements. Apart from diseased people, young and dynamic people like sportspersons often need replacements due to fracture and excessive strain. Especially after the world wars, the need for biomaterials was acutely felt and in the recent context of global terrorism, this field assumes much more significance.

The field of biomaterials is not new and as early as 4000 years back the Egyptians and Romans have used linen for sutures, gold and iron for dental applications and wood for toe replacement but with very little knowledge about the problem of corrosion. Nylon, Teflon, silicone, stainless steel and titanium were some of the other materials which were put into use after World War II. Currently, the availability of better diagnostic tools and advancements in the knowledge on materials as well as on surgical procedures, implantology has assumed greater significance and bioimplants are commonly used in dentistry, orthopedics, plastic and reconstructive surgery, ophthalmology, cardiovascular surgery, neurosurgery, immunology, histopathology, experimental surgery, and veterinary medicine (Fig. **1**). Various classes of materials such as metals, alloys, polymers ceramics and composites have been widely used to fabricate the bioimplants. These implants encounter different biological environments of very different physico-chemical nature and their interaction with the tissues and bones is a complex problem. Scientific knowledge was completely lacking in the early years of human existence and the credit for the origin and evolution of today's bioimplants are due to Harold Ridley, Paul Winchell, Per-Ingvar Branemark, Otto Wichterle, John Charnley and others. Their works at the laboratory were first tested on animals which led to the birth of the ultimate biomaterials that could be accepted by the human system.

The first and foremost requirement for the choice of the biomaterial is its acceptability by the human body. The implanted material should not cause any adverse effects like allergy, inflammation and toxicity either immediately after surgery or under post operative conditions. Secondly, biomaterials should possess sufficient mechanical strength to sustain the forces to which they are subjected so that they do not undergo fracture and more importantly, a bioimplant should have very high corrosion and wear resistance in highly corrosive body environment and varying loading conditions, apart from fatigue strength and fracture toughness. A biomaterial should remain intact for a longer period and should not fail until the death of the person. This requirement obviously demands a minimum service period of from 15 to 20 years in older patients and more than 20 years for younger patients. The success of a biomaterial or an implant is highly dependent on three major factors (i) the properties (mechanical, chemical and tribological) of the

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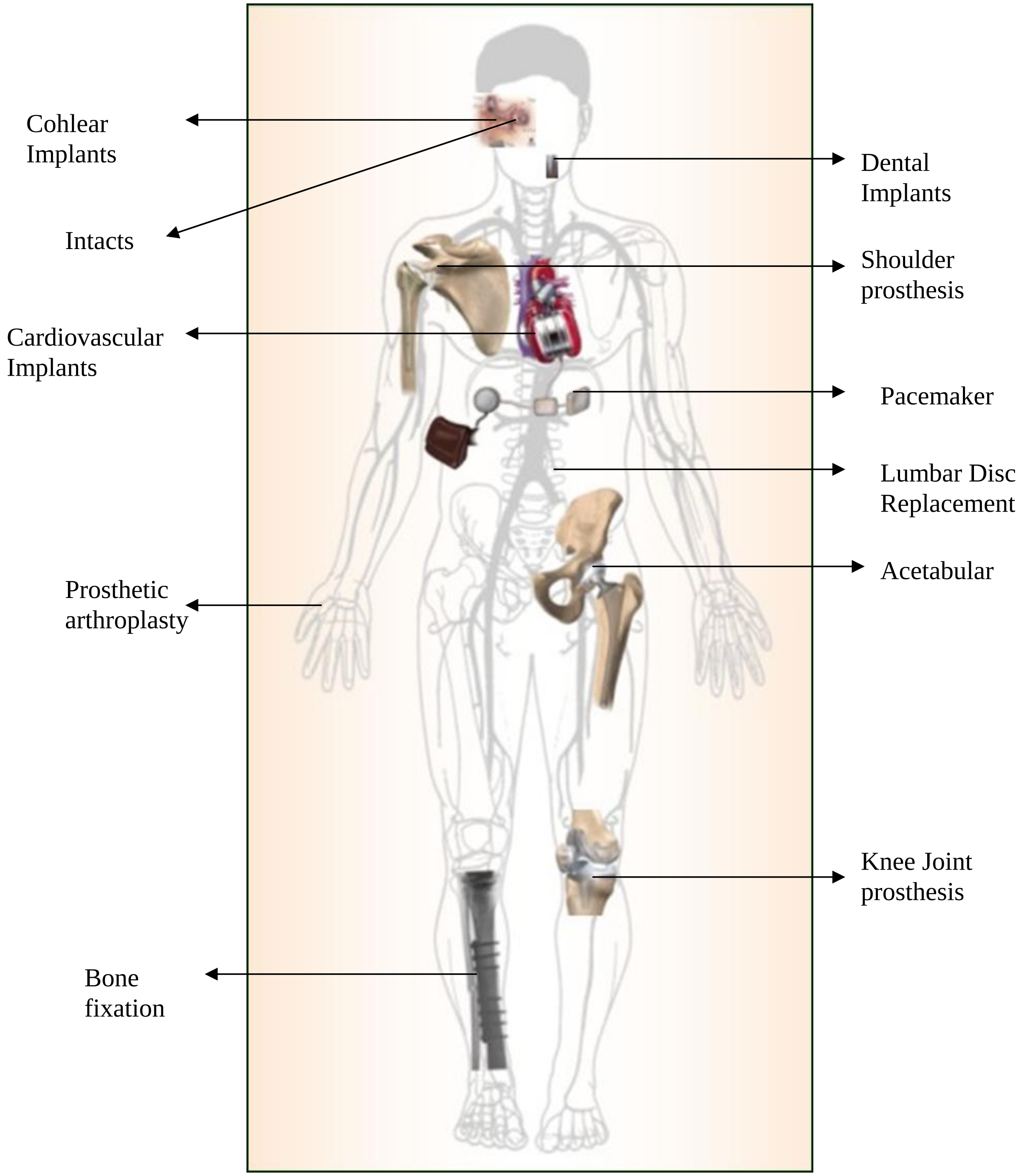


Fig. (1). Biomaterials for human application.

biomaterial in question (ii) biocompatibility of the implant and (iii) the health condition of the recipient and the competency of the surgeon. The currently used materials that were selected based on above mentioned criteria though function well in the human system are still found to generally fail within a period of about 12-15 years, which leads to revision surgery in order to regain the functionality of the system. The reasons for their failure are manifold which includes mechanical, chemical, tribological, surgical, manufacturing and biocompatibility issues. Out of all these issues, the failure of an implant due to corrosion has remained as one of the challenging clinical problems. This important field of research, over the years, has been discussed at length by several authors in the form of books [1-10] and comprehensive review articles [11-15] and the interested reader can go through them to gain mastery over this subject.

This article is divided into seven sections; section 1 discusses the reasons which lead to the corrosion of bioimplants in biological environment. The corrosion behavior of the surface oxide layer formed on the implants and the reasons for the failure of the implants are described in detail in sections 2 and 3 respectively. The corrosion behavior of conventional as well as the recently developed alloys are discussed at length in Section 4, whereas, Section 5 is devoted to a discussion on the corrosion of cardiovascular, dental and orthopedic implants. The prevention of corrosion by appropriate coating techniques is dealt with in section 6. The current and future developments with regard to the corrosion of biomedical implants are discussed in the final section 7.

1. WHY METALS CORRODE IN HUMAN BODY?

Corrosion, the gradual degradation of materials by electrochemical attack is of great concern particularly when a metallic implant is placed in the hostile electrolytic environment of the human body. The implants face severe corrosion environment which includes blood and other constituents of the body fluid which encompass several constituents like water, sodium, chlorine, proteins, plasma, amino acids along with mucin in the case of saliva [16]. The aqueous medium in the human body consists of various anions such as chloride, phosphate, and bicarbonate ions, cations like Na^+, K^+, Ca^{2+}, Mg^{2+} etc., organic substances of low-molecular-weight species as well as relatively high-molecular-weight polymeric components, and dissolved oxygen [17, 18]. The biological molecules upset the equilibrium of the corrosion reactions of the implant by consuming the products due to anodic or cathodic reaction. Proteins can bind themselves to metal ions and transport them away from the implant surface upsetting the equilibrium across the surface double layer that is formed by the electrons on the surface and excess cations in the solution. In addition, proteins that are absorbed on the surface are also found to reduce the diffusion of oxygen at certain regions and cause preferential corrosion at those regions. Hydrogen which is formed by cathodic reaction acts as a corrosion inhibitor, however, the presence of bacteria seems to change this behavior and enhance corrosion by absorbing the hydrogen present in the vicinity of the implant. Changes in the pH values also influence the corrosion. Though, the pH value of the human body is normally maintained at 7.0, this value changes from 3 to 9 due to several causes such as accidents, imbalance in the biological system due to diseases, infections and other factors and after surgery the pH value near the implant varies typically from 5.3 to 5.6. In spite of the fact that most of the materials used are protected by the surface oxide layers from the environmental attack, there is clinical evidence for the release of metal ions from the implants and this leaching has been attributed to corrosion process.

It has been well accepted that the tolerable corrosion rate for metallic implant systems should be about 2.5 x 10^{-4} mm/yr, or 0.01 mils/yr [19]. The most common forms of corrosion that occur are uniform corrosion, intergranular, galvanic and stress corrosion cracking, pitting and fatigue corrosion. Even though new materials are continuously being developed to replace implant materials used in the past, clinical studies show that these materials are also prone to corrosion to a certain extent [20]. The two physical characteristics which determine implant corrosion are thermodynamic forces which cause corrosion either by oxidation or reduction reaction and the kinetic barrier such as surface oxide layer which physically prevents corrosion reactions [20-22].

In some cases though the material will not fail directly due to corrosion, it is found to fail due to accelerated processes such as wear and fretting leading to tribocorrosion. Fretting results in the rupture of protective oxide layer, initiation of cracks and formation of reactive metal atoms on the surface that are susceptible to corrosion [23]. In order to limit further oxidation, initially formed passive films must have certain characteristics; i) non - porous ii) atomic structure that will limit the migration of ions and electrons across the metal oxide - solution interface and iii) high abrasion resistance. Hence, when a material is developed for implant application, it should not only be subjected to basic corrosion screening test, but also has to be tested for its behaviour under different conditions such as reciprocatory wear, fretting, stress corrosion etc depending up on their applications. There are ASTM standards for testing corrosion resistance of these materials under different conditions. The commonly used standards for testing different corrosion processes are given in Table **1** [24]. Corrosion is accelerated in the presence of wear and also simultaneous corrosion and wear are often encountered in biomedical implants. Dearnley *et al.* have evaluated the corrosion behavior of the scratched coated specimens to determine the wear accelerated corrosion behavior of the coatings and also suggested a methodology to measure the simultaneous corrosion and wear of a material [25]. However it is important to note that there are no standards available to test the tribocorrosion behavior of the implants.

Table 1. Standards for Testing Corrosion Resistance of Biomaterials

ASTM Standards	Specifications
ASTM G 61-86, and ASTM G 5-94	Corrosion performance of metallic biomaterials
ASTM G71-81	Galvanic corrosion in electrolytes
ASTM F746-87	Pitting or crevice corrosion of metallic surgical implant materials
ASTM F2129-01	Cyclic potentiodynamic polarization measurements

2. SURFACE OXIDE FILM ON METALLIC MATERIALS IN BIOLOGICAL ENVIRONMENT

Surface oxide film formed on metallic materials plays an important role as an inhibitor for the release of metallic ions and the behavior of the surface oxide changes with the release of ions. Further, the composition of the surface oxide film changes according to reactions between the surfaces of metallic materials and living tissues. Even low concentration of dissolved oxygen, inorganic ions, proteins, and cells may accelerate the metal ion release. In addition, the dissolution of surface oxide film due to active oxygen species has also been reported [26]. The regeneration time of the surface oxide film after disruption also decides the amount of ions released. Tissue compatibility, the prerequisite for an implant is basically determined by the nature of the reactions which take place at the initial stages after implantation and thus the success of the implant depends on the reactions taking place between the surface of metallic materials and living tissues soon after the fixation of the implant. Surface oxide films present on metallic materials play a very important role, not only for corrosion resistance but also for tissue compatibility. Therefore, as pointed out by Kasemo and Lausma [26], it is important to analyze the surface characteristics of these materials when discussing the issues of corrosion and tissue compatibility. Biomedical implants should be subjected to both *in vitro* and *in vivo* studies for their applications. *In vitro* studies which are performed in simulated body condition give an overview of the behavior of the material under the given condition and obviously it cannot be taken as the final test to recommend a material as an implant. The *in vivo* tests which are performed using animal models evaluate the actual performance of the materials and these tests are required in order that it is approved by FDA (Food and Drug Administration, USA). *In vitro* corrosion studies on orthopaedic biomaterials are carried out either in Hank's solution or Ringer' s solution whose constituents are given in Tables **2** and **3** respectively, whereas the corrosion resistance for dental materials is evaluated using synthetic saliva whose constituents could be seen in Table **4**. It should be mentioned here that various compositions have also been suggested which are close to the natural saliva [27]. Three of the existing saliva substitutes are xialine 1, xialine 2 and saliveze, where xialine 1 and xialine 2 are based on xanthan gum and the saliveze is based on carboxymethylcellulose. Main constituents of artificial saliva are Mg^{2+}, K^+, Na^+, Cl^-, SCN^-, NH_4^+, Ca^{2+}, CO_{3tot} (where, $[CO_3]_{tot} = [CO_3^{2-}] + [HCO_3^-]$), and the pH is near neutral. In addition to these ions, presence of fraction of organic compounds such as glycoprotein, have been reported in the saliva and it plays an important role in maintaining the viscosity which, in turn, affects the diffusion of various ions. However, its effect on the corrosion of biomaterials remains to be understood [28]. The oxide film which inhibits the dissolution of metal ions is not always stable in the human body and hence a thorough understanding of the behavior of the oxide film in *in vivo* condition is essential to have a better insight of the corrosion phenomenon. The analysis of the surface oxide film on various metallic biomaterials is given in Table **5**.

Table 2. Composition of Hank's Solution

Substance	Composition (g L^{-1})
NaCl	8.0
KCl	0.4
$NaHCO_3$	0.35
$NaH_2PO_4.H_2O$	0.25
$Na_2HPO_4.2H_2O$	0.06
$CaCl_2$.2H2O	0.19
MgCl2	0.19
$MgSO_4$.7H2O	0.06
glucose	1.0
pH	6.9

[Ref: Bundy KJ. Corrosion and other electrochemical aspects of biomaterials. Crit Rev Biomed Eng 1994; 22: pp. 139-251].

Table 3. Composition of Ringer's Solution

Substance	Composition (g L^{-1})
NaCl	8.69
KCl	0.30
$CaCl_2$	0.48
pH	6.4

[Ref: Gonzalez EG, Mirza-Rosca JC. Study of the corrosion behavior of titanium and some of its alloys for biomedical and dental implant applications. J Electroanal Chem 1999; 471: p. 109].

Table 4: Composition of Different Artificial Saliva

Components	Artificial Saliva		
	Xialine1 (g L^{-1})	Xialine2 (g L^{-1})	Saliveze (g L^{-1})
Xanthan gum	0.92	0.18	-
Sodium carboxymethylcellulose	-	-	10
Potassium chloride	1.2	1.2	0.62
Sodium chloride	0.85	0.85	0.87
Magnesium chloride	0.05	0.05	0.06
Calcium chloride	0.13	0.13	0.17
Di-potassium hydrogen orthophosphate	0.13	0.13	0.80
Potassium di-hydrogen orthophosphate	-	-	0.30
Sodium fluoride	-	-	0.0044
Sorbitol	-	-	29.95
Methyl p-hydroxybenzoate	0.35	0.35	1.00
pH	Neutral	Neutral	Neutral

[Ref: Preetha A, Banerjee R. Comparison of artificial saliva substitutes. Trends Biomater Artif Organs 2005; 18(2): pp. 178-186].

When the surface oxide film of a metallic material is disrupted, corrosion proceeds and metal ions are released continuously unless the film is regenerated. The interactions between the physiological medium and the material play a decisive role on the reformation of the oxide layer and the time taken for the same. The time taken for repassivation which is also termed as regeneration time is different for various materials used. The corrosion rate following the disruption and the quantity of released metal ions depend up on the above said regeneration time. Regeneration time taken to form surface oxide films for various alloys is illustrated in Fig. (**2**). From these observations, it is found that the regeneration time is longer in stainless steel and shorter in Ti-6Al-4V, an alloy which is well known and widely used for orthopedic applications, indicating a fact that larger number of metal ions being released from stainless steel compared to the latter, which brings out one of the superior qualities possessed by this alloy, apart from its other advantageous properties. The repassivation rate of Ti in Hank's solution is found to be slower than that in saline and remains uninfluenced by the pH of the solution. In addition, the surface oxide film regenerated on Cp Ti in Hanks' solution contains phosphate ions in the outer layer. Phosphate ions are preferentially filled up during regeneration of surface oxide film on titanium and the film

Table 5. Analysis of the Surface Oxide Film on Various Metallic Biomaterials

Metallic Biomaterial	Surface Oxides	Surface Analysis
Titanium(Ti)	Ti^{0+}, Ti^{2+}, Ti^{3+}, Ti^{4+}	• Ti^{2+} oxide thermodynamically less favorable than Ti^{3+} formation at the surface. • Ti^{2+} and Ti^{3+} oxidation process proceeds to the uppermost part of the surface film and Ti^{4+}observed on the surface outer most layer.
Titanium alloys		
Ti-6Al-4V	TiO_2	Surface consists of small amount of Al_2O_3, hydroxyl groups, and bound water and the alloying element Vanadium was not detected
Ni-Ti	TiO_2-based oxide	Minimal amounts of nickel in both oxide and metal states
Ti-56Ni	TiO_2	Very low concentrations of metallic nickel, NiO, hydroxyl groups and bound water on the surface were detected.
Ti-Zr	Titanium and Zirconium oxides	Titanium and zirconium are uniformly distributed along the depth direction. The thickness of the oxide film increases with increase in zirconium content.
Stainless steel		
Austenitic stainless steel	Iron and chromium	Only very less amount o f molybdenum was observed on the surface and nickel was absent when tested in both the air and in chloride solutions.
316L	Oxides of Iron, chromium, nickel, molybdenum and manganese(thickness about 3.6 nm)	The surface film contains a large amount of OH^-, that is, the oxide is hydrated or oxyhydroxide. Iron is enriched in the surface oxide film and nickel, molybdenum, and manganese are enriched in the alloy substrate just under the surface oxide film.
Co-Cr-Mo alloy		
Co-36.7Cr-4.6Mo	Oxides of cobalt and chromium without molybdenum(thickness 2.5 nm)	Surface contains large amount of OH^-, that is the oxide is hydrated or oxyhydroxidized. Chromium and molybdenum distributed more at the inner layer of the film.

consists of titanium oxide and titanium oxyhydroxide containing titanium phosphate. Calcium and phosphate ions are also adsorbed to the film after regeneration, and calcium phosphate or calcium titanium phosphate are formed on the outermost surface. In Ti-6Al-4V also calcium phosphate was observed on the surface oxide film regenerated in Hanks' solution and on other hand only phosphate without calcium is formed on Ti-56Ni, Ti-Zr and Zr based biomaterials. Thus the composition of surface oxide layer and its interaction with the environment is highly dependent on the constituents of the material used. The stability of the surface oxide layer in 316L SS as well as in Ni-Ti is not very high and the possibility of metal ions being released is greater when compared to conventional alloys such as Co-Cr and Ti-6Al-4V. Hence, in general a coating given on the implants is preferable as it will reduce the corrosion rate.

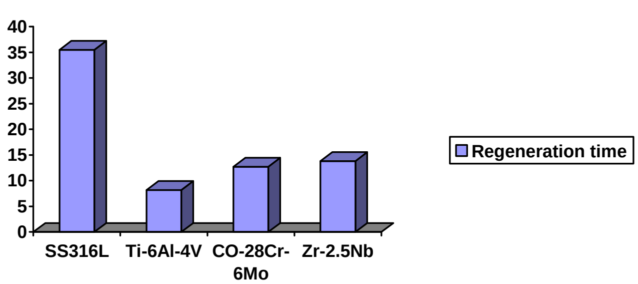


Fig. (2). Regeneration time of surface oxide films for various alloys. [Ref: Hanawa T. Reconstruction and regeneration of surface oxide film on metallic materials in biological environments. Corrosion Rev 2003; 21: pp. 2-3].

3. EFFECT OF CORROSION-FAILURE OF IMPLANTS

The reaction of the metallic ions that leaches away from the implant due to corrosion in the human body affects several biological parameters. As a material starts to corrode, the dissolution of metal will lead to erosion which in turn will eventually lead to brittleness and fracture of the implant. Once the material fractures, corrosion gets accelerated due to increase in the amount of exposed surface area and loss of protective oxide layer. If the metal fragments are not surgically extracted, further dissolution and fragmentation can occur, which may result in inflammation of the surrounding tissues. Table **6** illustrates the effects of corrosion in human body due to various biomaterials, whereas Table **7** shows the types of corrosion in the conventional materials. The contents of the Table **6** amply illustrate the possible hazardous effects associated with the corroded implant material. The release of corrosion products will obviously lead to adverse biological reactions in the host, and several authors have reported increased concentrations of corroded particles in the tissue near the implants and other parts of the human body such as kidney, liver etc. [29, 30]. In spite of the fact that there is no histological evidence to show the slow release of metallic ions due to corrosion, the discoloration of the surrounding tissue and the foreign body reactions clearly indicate that this is due to corrosion of implants [31].

Cobalt-chromium alloy which is a commonly used biomaterial consists of the elements cobalt, chromium, nickel and molybdenum. It is felt that the corrosion of cobalt-chrome in the wet and salty surroundings of the human body, releases toxins into the body which in turn leads to the formation of cancerous tumors. Though the number of tumors near the implant formed may be less, there is a possibility that many could exist at other parts of the human body due to the released ions. Hence there is always a need to develop new and safer materials which have extremely high corrosion resistance.

Table 6. Effects of Corrosion in Human Body Due to Various Biomaterials

Biomaterial Metals	Effect of Corrosion
Nickel	Affects skin - such as dermatitis
Cobalt	Anemia B inhibiting iron from being absorbed into the blood stream
Chromium	Ulcers and Central nervous system disturbances
Aluminum	Epileptic effects and Alzheimer's disease
Vanadium	Toxic in the elementary state

[Ref: Aksakal B, Yildirim ÖS, Gul H. Metallurgical failure analysis of various implant materials used in orthopedic applications. J Fail Anal Prevent 2004; 4(3): p. 17].

Aksakal *et al*. investigated failed implants made of titanium alloy Ti-6Al-4V and 316L SS that were removed from several patients [32]. The failure analysis studied using scanning electron microscope (SEM) is represented in Fig. (**3**). From the analysis it was evident that the failure of femoral titanium plates has occurred through corrosion fatigue which was promoted by the presence of intense localized corrosion and intergranular cracking. In addition, corrosion fatigue and fretting corrosion have also been observed in bone plates and screws at the bone-stem, and stem-cement interfaces of modular hip implants. The failure of stent device due to stress corrosion fatigue is a long term problem and this failure has been attributed to the weak surface of the implant [33]. This stress corrosion cracking in biomedical implants can lead to loss of structural integrity of the implanted device and its functions. Thus these complications lead to disintegration of the implant [34]. It has been found that this dissolution of metal ions can be reduced by suitable biocompatible inorganic coatings, such as hydroxy-apatite (HAP) coating with some binders, and this can lead to delay in corrosion and wear and also minimize the loosening of implants from bone [35]. Thus the only solution to impede corrosion is by choosing better quality materials with appropriate coating.

4. CORROSION OF CONVENTIONAL ALLOYS

The commonly used surgical implants are usually made from one of the three types of materials: austenitic stainless steel, cobalt-chromium alloy, and titanium and its alloys and out of these, 316L austenitic stainless steel is the most commonly used implant material as it is cost effective [36].

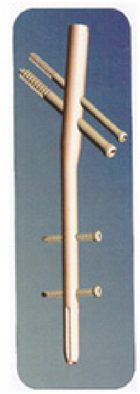
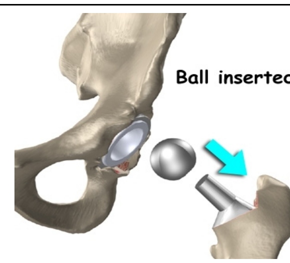
4.1. Austenitic Stainless Steels

The most commonly employed steel alloys are 316 and 316L grades. ASTM recommends type 316L for implant fabrications for the obvious reason that presence of less carbon decreases the chance of forming chromium carbide that generally results in intergranular corrosion. Lowering of the carbon content also makes this type of stainless steel more corrosion-resistant to chlorine-bearing solutions such as physiological saline in the human body [4]. However stainless steel is susceptible to localized corrosion by chloride ions and reduced sulfur compounds [37]. The presence of micro organisms on a metal surface often leads to highly localized damages in the concentration of the electrolytic constituents, pH and oxygen levels [38]. Studies on corrosion and electrochemical behavior of 316L SS in the presence of aerobic iron-oxidizing bacteria (IOB) and anaerobic sulfate-reducing bacteria (SRB) reveal that the interactions between the stainless steel surface with the corroded products, bacterial cells and their metabolic products increases the corrosion damage and also accelerates pitting propagation [39]. In this respect, the decreasing antibacterial activity exhibited by different materials is given in the following order gold > titanium > cobalt > vanadium > aluminum > chromium > iron [40]. Studies on retrieved implants show that more than 90% of the failure of implants of 316L SS are due to pitting and crevice corrosion attack [41]. These localized corrosion attacks and leaching of metallic ions from implants necessitate improvement in the corrosion resistance of the currently used type 316L SS by bulk alloying or modifying the surface [42].

Biomedical materials which are subjected to cyclic loading and high stresses in the presence of aggressive environment fail due to fatigue [4, 6, 26]. Fatigue process is found to get further accelerated due to the formation of wear debris leading to fatigue wear. During fatigue there is disruption of the oxide layer and the inability of the material to repassivate immediately exposes some region of the metal to the environment leading to corrosion. The initiation of crack due to fatigue was observed during the measurement of corrosion potential of cold worked 316L SS and it was also observed that the fatigue strength dropped drastically when the repassivation was suppressed, thus, confirming that oxide layer formation plays a vital role in the determination of the fatigue life of the materials exposed to aggressive corrosive environment [4]. In addition, fretting that occurs between implant and bone is also found to accelerate the fatigue as the repassivation becomes more difficult in the presence of fretting.

Williams, Sivakumar *et al.* and Frazad *et al.* have extensively reviewed the failure of stainless steel implants [1, 36, 43]. The studies which have been made by Farzad *et al.* on the stainless steel implants that were fractured in patients thighs revealed interesting results. Several damage mechanisms such as crevice corrosion, pitting, initiation of cracks from these pits, intergranular surface cracking inside the crevice, and also stress corrosion cracking (SCC)-like branched cracks where observed in the failed implants. But, the main failure mechanism was determined to be corrosion fatigue assisted by crevice corrosion. Apart from the intensive weakness of the alloy against the crevice corrosion, the sulphide inclusions had further assisted the formation of the corrosion pits in the crevice regions. These results were further corroborated by the observations made by Sivakumar

Table 7. Types of Corrosion in the Conventional Materials Used for Biomaterial Implants

Type of Corrosion	Material	Implant Location	Shape of the Implant
Pitting	304 SS, Cobalt based alloy	Orthopedic/ Dental alloy	
Crevice	316 L stainless steel	Bone plates and screws	
Stress Corrosion cracking	C0CrMo, 316 LSS	Only in *in vitro*	
Corrosion fatigue	316 SS, CoCrNiFe	Bone cement	
Fretting	Ti6Al4V, CoCrSS	Ball Joints	
Galvanic	304SS/316SS, CoCr+Ti6Al4V, 316SS/Ti6Al4V Or CoCrMo	Oral Implants Skrews and nuts	
Selective Leaching	Mercury from gold	Oral implants	

[Ref: Blackwood DJ. Biomaterials: past successes and future problems. Corrosion Rev 2003; 21(2-3): pp. 97-124.]

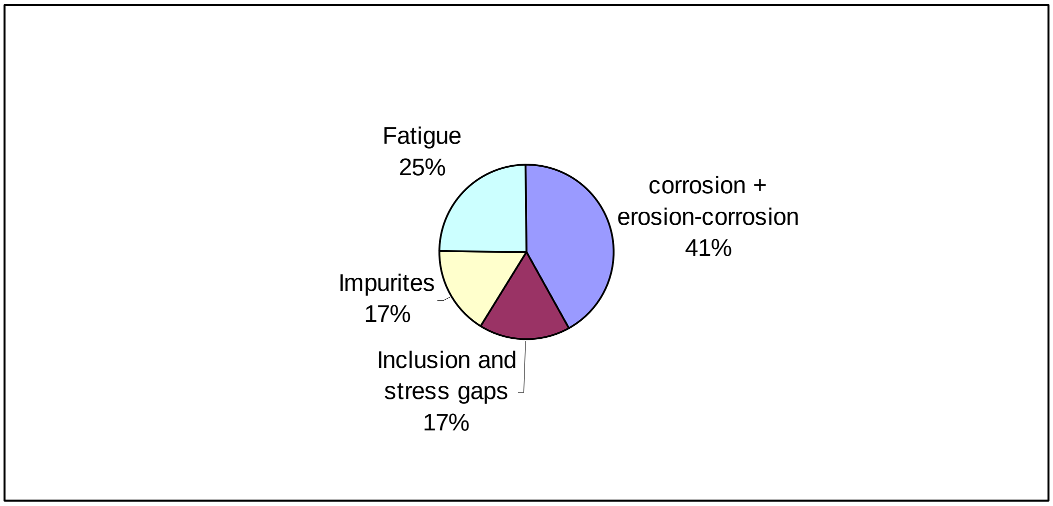


Fig. (3). Failure analysis of implants Ti6Al4V and 316L steel. Ref: Aksakal B, Yildirim ÖS, Gul H. J Fail Anal Prevent 2004; 4(3): p. 17].

et al. on several patients implanted with stainless steels. The studies revealed that the failures are due to various corrosion mechanisms and the percentage of corrosion in various anatomical positions are illustrated in Figs. (**4a**, **4b**) [41]. These studies suggested that improvement in the design of the implants could reduce the number of metal-metal interfaces or reduction of the crevice area could prevent crevice corrosion. In addition to crevice corrosion, pit-induced fatigue failure was also observed in the compression bone plate and pit induced stress corrosion cracking in the intramedullary nail. The corrosion failure not only impairs the performance of the permanent implants but also the behavior of the temporary implants made of surgical grade type 316L SS [44].

4.2. Cobalt-Based Alloys

These alloys have better mechanical strength, elastic modulus, abrasion resistance and corrosion resistance compared to that of stainless steel. As with stainless steel, chromium in these alloys provides the essential corrosion resistance. But in contrast to that of the stainless steel, cobalt also contributes to the corrosion resistance and this makes cobalt-chromium based alloys to have excellent corrosion resistance [45]. Because of their outstanding mechanical properties, apart from these alloys being used for the fabrication of removable partial dentures, they are also used for making implants that require fine framework constructions [46]. Cobalt-based alloys have been widely employed in orthopaedic implants and biocorrosion of this alloy is one of the major problems to be dealt with as there is larger release of metal ions which causes adverse effects [47]. Co-Cr-Mo alloy is used as a femoral head of joint prostheses in conjunction with an ultra high molecular weight polyethylene (UHMWPE) cup because of the high wear and corrosion resistance of this alloy. The problem with the metal-on-metal couple is that the release of metal ions is higher than that of the polymer-on-metal couple in *in vivo*, which will, over many years lead to toxicity problem. The metal ions dissolved from Co-Cr-Mo alloy powder bind serum proteins to a much greater level compared to that of Ti-6Al-4V alloy. The conventional Co-29Cr-6Mo-1Ni alloy (ASTM F75-92) contains 1 mass% of Ni and the Ni and Co ions are responsible for allergic reactions. Especially, Ni causes carcinogenicity and the decrement of Ni from Co-Cr-Mo alloy is one of the solutions for the toxicity problem of cobalt nickel based systems [48].

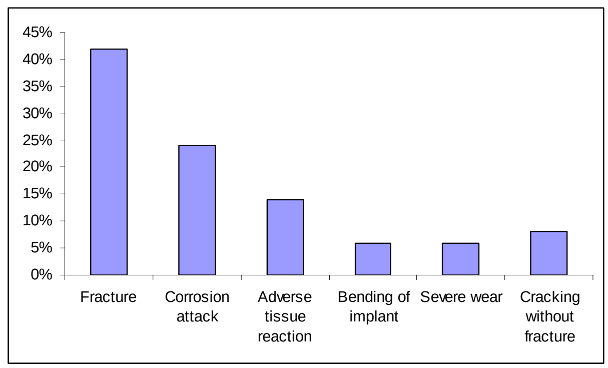


Fig. (4a). Causes for failure of SS implants.

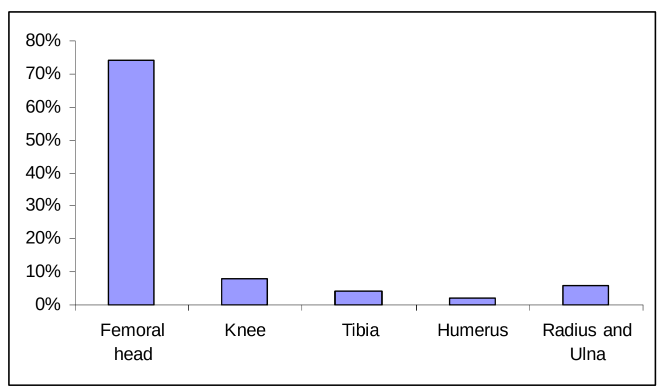


Fig. (4b). Failure of SS implants at various anatomical locations (Ref. Sivakumar M, Suresh Kumar Dhanadurai K, Rajeswari S, Thulasiraman V. Failures in stainless steel orthopaedic implant devices: a survey. J Mater Sci Lett 1995; 14: pp. 351-4).

4.3. Titanium-Based Alloys

Since 1970s the application of titanium and its alloys have become more widespread as they possess high strength, low modulus, lower density, and a good combination of mechanical and outstanding corrosion resistance [49]. In general more than 1000 tonnes (2.2 million pounds) of titanium devices are implanted in patients worldwide every year and also the medical grade titanium alloys have a significantly higher strength to weight ratio than competing stainless steels. It has been well established that titanium is completely inert and immune to corrosion by all body fluids and tissue and is thus completely biocompatible [50]. High modulus of elasticity of the conventional alloys has resulted in the stress shielding effect and the failure of the implant. The modulus of elasticity of titanium based alloys is much lower and closer to that of the bone when compared to SS and Co-Cr alloys and hence they are more preferred for long term applications. As of now, they are used as implants for joint replacements, bone fixation, dental implants, heart pacemakers, artificial heart valves, stents and components in high-speed blood centrifuges because of their high specific strength and chemical stability [12]. However, these implants such as artificial joints and bone plates are likely to be damaged mostly due to fatigue [51]. The reason for this is due to the decrease in fatigue strength, which in turn should arise from the synergistic effect of the formation of corrosion pits on the surface, which arise from the dissolution of Ti^{2+} ions in the living body, wearing at sliding parts and fretting [52, 53].

Although Ti-6Al-4V alloy has got several positive features, detailed studies have shown that they lead to long term ill effects such as peripheral neuropathy, osteomalacia and Alzheimer disease due to the release of aluminum and vanadium ions from the alloy. In addition to this, vanadium which is present both in the elemental state and in oxides (V_2O_5) is also toxic [54]. Further, Ti-6Al-4V alloy has a lower wear resistance and higher elastic modulus than bone which leads to ''stress shielding effect'' [55, 56]. Corrosion fatigue could occur in titanium hip implants by even simple walking that causes cyclic loading at about the frequency of 1 Hz. Fatigue corrosion resistance of titanium is almost independent of the pH value while the fatigue corrosion strength of stainless steel starts decreasing below a pH value of 4. The study of Yu *et al*. reveals that the nitrogen ion implantation and heat treatment procedures enhance the corrosion fatigue of Ti-6Al-4V alloy [57]. The clinical concern with titanium and its alloys is because of its known potential toxicities associated with the alloying elements and known pathologies due to the particles which emanate due to the breakdown of oxide layer associated with metal implant degradation. In the case of Ti-6Al-4V alloy, vanadium oxide in the passive film dissolves and results in the generation and diffusion of vacancies in the oxide layer [58]. However, the addition of alloying element such as Nb enhances the passivation by the formation of Nb rich pentoxide which is highly stable in the body environment leading to high corrosion resistance [59]. A comparative study on the corrosion behavior between Ti-Ta and Ti-6Al-4V alloys showed that the addition of Ta greatly reduces the concentration of metal release from the surface oxide layer because of the formation of highly stable Ta_2O_5 oxides [60]. Thus the corrosion resistance of the passive film is very much dependent on the thickness of the layer formed and the nature of the elements present in titanium and its alloys [14].

The corrosion of NiTi alloys used for dental, orthopedic and cardio vascular applications is debatable as there are conflicting reports on their corrosion resistance when compared to Cp Ti or Ti-6Al-4V. Many studies have shown that NiTi is highly compatible with living tissues but adverse effects caused by this material are also often reported. In particular, severe cell death arising from the poor corrosion resistance and toxic constituents such as Ni in NiTi alloys has been observed [61]. Though this alloy exhibits high corrosion resistance to pitting tested at passive conditions and also potentiodynamic test at normal ranges of pH and temperatures corresponding to human body, there are reports which show the inferior corrosion behavior of this alloy when compared to Ti-6Al-4V and also stainless steel based materials when measured using potentiostatic scratch test method. This clearly indicates that the passive layer formed on NiTi is less protective than that on Ti-6Al-4V. Oxide films present on different metal surfaces are characterized by electrochemical impedance spectroscopy (EIS). Titanium and its alloys have been analyzed extensively by EIS in different media such as in Hank's solution [62], Ringer solution [15], 0.9% NaCl [63], 10% KCl and 30% KCl solutions [64], phosphate buffer solution (PBS) and artificial saliva [65]. However, research work carried out on the interactions between the materials and biological systems in *in vivo* are relatively new and not yet advanced. Hence a systematic study based on physical chemistry and life science is required to understand the formation of the oxide film and repassivated layer obtained under different environments.

4.4. Magnesium and its Alloys

Recently magnesium alloys are emerging as a better alternative for temporary implants and for making stents [5]. As the currently used permanent cardiovascular implants pose several problems such as thrombogenicity, permanent physical irritation, mismatches in mechanical behavior between stented and non-stented vessel area etc., a new domain of research on metallic implants focuses on new biodegradable implants, which dissolve in biological environment after a certain time of functional use. The degradable implants are seen to provide more physiological repair, better reconstruction, appropriate radial and longitudinal straightening effect and tissue growth. Magnesium as a degradable implant material provides both biocompatibility and suitable mechanical properties in *in vitro* and *in vivo* studies and it has been shown that magnesium is suitable to be used for degradable implants as they exhibit good cell attachment and tissue growth [5, 66, 67]. However the studies concerning the corrosion behaviour of magnesium in biological environment and cytotoxicity of magnesium are lacking. Recently, a number of studies have been carried out to investigate the corrosion behaviour of magnesium alloys in artificial physiological fluids and most of them are Al containing Mg alloys [68]. Most alloying elements will dissolve into the human body during the magnesium alloy degradation, for example, when the AZ91D magnesium alloy is used, the Al present in the alloy would get into the human body, which might be hazardous, from the health point of view. The magnesium alloy implants are expected not to degrade until the healing is completed and tissue growth has occurred. However it is unfortunate to observe that magnesium and its alloys corrode too quickly at the physiological pH of 7.2 to 7.4 as well as in physiological media containing high concentrations of aggressive ions, thereby loosing mechanical integrity before the tissues have sufficient time to heal [69]. Various methods such as alkali heat treatment [70], plasma immersion ion implantation (PIII) [71], micro-arc oxidization (MAO) [72], and anodic oxidization [73] have been proposed to improve the corrosion resistance of Mg alloys. Thus Mg systems open a new window in the field of cardio vascular implants.

5. CORROSION OF VARIOUS IMPLANTS

5.1. Cardiovascular Implants

Heart diseases and especially the ischaemic heart disease, causes yearly deaths of about 180,000 people in UK and of more than 500,000 in USA and this is the main reason for the premature death of middle-aged men and women [74, 75]. Artificial heart is considered to be a solution for this and they are made up of a pair of substantially seamless, polyurethane rigid outer housings each having a shape approximating the combined outer shape of a natural cardiac ventricle and its associated auricle, with the bases of each of the housings being bounded together (Fig. **5**) [76]. Cardiovascular implants should possess unique blood biocompatibility to ensure that the device is not rejected due to adverse thrombogenic (clotting) or hemodynamic blood responses. Though cardiovascular implants can be fabricated using natural tissues, the gradual calcification of this bioprostheses leading to the eventual stiffening and tearing of the implant is of clinical apprehension. Non-bioprosthetic implants are fabricated from materials such as pyrolytic carbon-coated graphite, pyrolytic carbon coated titanium, stainless steel, cobalt-chrome alloys, cobalt-nickel alloys, alumina coated with polypropylene and Poly-4-fluoroethylene [77]. Apart from the limitations on materials imposed by the requirements of blood biocompatibility and the problems associated with designs imposed by the need to optimize blood flow, proper care should be taken in all respects in order to avoid the risk of a second surgical procedure. Further, if there is a catastrophic failure of the implanted device, it will certainly result in the death of the patient [78].

316 SS, Ti and its alloys (shape memory alloys), Co-Cr are the most frequently used metallic materials for stents. These alloys are prone to various extents of corrosion. Corrosion of bare metal stents is illustrated in Fig. (**6**). 316L SS is the most commonly used as a metal for stents either with or without coating material. Allergic reactions caused by the release of nickel have been found to occur among SS implants. In particular, the release of nickel, chromate, and molybdenum ions from SS stents may trigger local immune response and inflammatory reactions. Co-Cr alloys, which conform to ASTM standards F562 and F90, have been used in dental and orthopedic applications for the past few decades [79] and recently these alloys are being used for making stents because of their high elastic modulus (210 GPa). In general, Ni-Ti alloy constituting 49.5 - 57.5 at% nickel and the remaining Ti is widely used for fabricating self-expanding stents mainly because of its shape memory effect. Self-expanding stents have a smaller diameter at room temperature and capable of expanding up to their preset diameter at body temperature. After implantation it regains its original shape and conforms to the vessel wall because of the increase in temperature inside the body [80]. Biocorrosion of magnesium based alloys is a new area of study for improving cardiovascular implant as an effective temporary system with inherent or hybrid local drug delivery functions. Though degradable stents seem to offer an ideal solution for the corrosion of stainless steel device, it is difficult to adapt these materials as there are many factors that govern the success of the implant which remain unclear and further research is required in this area.

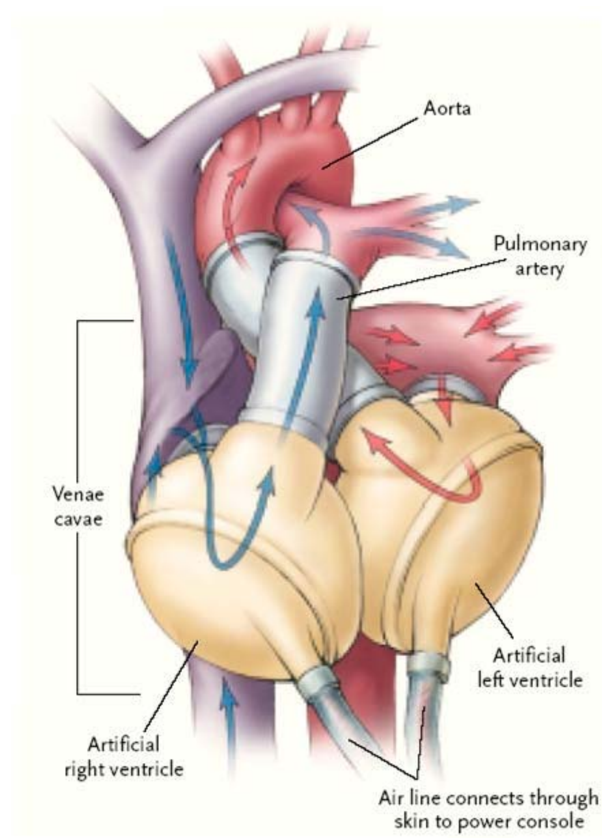


Fig. (5). Illustration of Artificial heart. [Ref: www.medgadget.com/archives/img/131113.jpg].

5.2. Corrosion of Dental Implants

The major classifications of dental implants are endosseous implants that are placed within the bone and the subperiosteal implants are placed on the top of the bone. Endosseous implants are further subdivided into two groups; root and blade forms and these types are used to secure either single crowns, fixed bridges or to retain removable prosthesis (dentures). As mentioned earlier, these implants face very aggressive environment in the mouth, the pH of saliva varies from 5.2 to 7.8. Thus the major reasons for corrosion of metallic implants and fillings are temperature, quantity and quality of saliva, plaque, pH, protein, and the physical and chemical properties of food and liquids as well as oral health conditions. Chaturvedi has reviewed extensively the corrosion aspect of dental implants [13]. As two metallic components are used together in making dental implants, galvanic corrosion occurs very frequently in dental implants. This occurs most commonly between the pair of metallic implants such as Co-Cr alloys, Ni-Cr, silver-palladium, gold and Ternary Ti dental implants. Pitting corrosion of cobalt based alloys leads to the release of

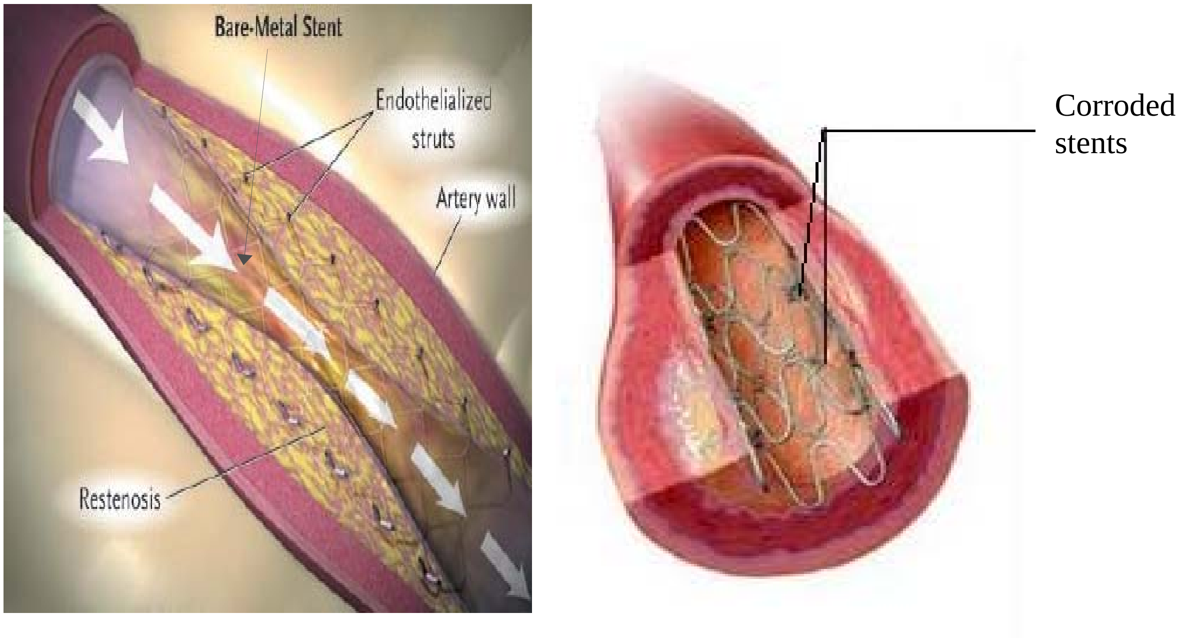


Fig. (6). Corrosion of stent in bare metal.

carcinogens into the body [81]. On the other hand titanium and its alloys are highly resistant to pitting corrosion in different *in vivo* conditions encountered; however they undergo corrosion in high fluoride solutions in dental cleaning procedures [82]. The corrosion products cause discoloration of the adjacent soft tissue, allergic reactions and rashes in some patients. The wound healing process is also found to be modulated by the metal ions released by corrosion. The corrosion of the implants is further accelerated in the absence of poor osseointegration. It is Branemark who coined the term osseointegration with respect to dental implants and worked extensively on this process of integration of metallic implant with surrounding bone [83].

As gold, palladium, and platinum are chemically stable and do not undergo significant corrosion they are highly preferred, however, the cost constraints have led to the use of implants that are made up of different alloys hitherto discussed. Ti and its alloys are more preferred because of the reasons which have been described at length in the earlier sections. Pure titanium castings have mechanical properties similar to gold alloys and some titanium alloy castings, such as Ti-6Al-4V and Ti-15V have properties closer to Ni-Cr and Co-Cr castings. Though materials may exhibit high corrosion resistance in *in vitro* conditions they are found to fail in clinical conditions as they face extreme conditions such as change in saliva composition, oral hygiene, diet, variation in stresses and brushing methodology. Thus it is essential a material is screened for corrosion behavior in all extreme conditions such that it does not fail in actual applications.

5.3. Corrosion of Orthopedic Implants

Orthopedic implants include both temporary implants such as plates and screws and permanent implants that are used to replace hip, knee, spinal, shoulder, toe, finger etc. The corrosion mechanisms that occur in temporary implants are crevice corrosion at shielded sites in screw/plate interface and beneath the heads of fixing screws and pitting corrosion of the implants made of SS [84, 85]. The main cause for the failure of the orthopedic implants is wear, which in turn is found to accelerate the corrosion. Hence, high wear resistant materials such as ceramics, Co-Cr are often preferred to fabricate orthopedic implants. In hip implants, Ti based alloys are used only for making the femoral component and the ball is either made of Co-Cr or other hard ceramics. The femoral components are sometimes coated with cement to have good fixation. Waller *et al.* have observed crevice corrosion in femoral components made of Ti-6Al-4V and Ti-6Al-7Nb when they were implanted with bone cement [86]. *In vitro* studies on the corrosion behavior of various titanium alloys by Nakagawa *et al.* revealed that titanium with Pd exhibited high resistance to corrosion over a wide range of pH due to the enrichment of palladium on the surface [87]. The accelerated corrosion test performed by Khan *et al.* on Cp Ti, Ti-Nb-Zr and Ti-Mo alloys in *in vitro* conditions demonstrated Ti-6Al-7Nb and Ti-6Al-4V possessed best combination of corrosion and wear [88]. However, the nature and distribution of corrosion products released into the body from these orthopaedic implants remains, still as an important issue [89]. Hence, currently several researchers are working on the enhancement on the improvement of surface properties of titanium based alloys [14, 42].

6. SURFACE MODIFICATION IN BIOMATERIALS FOR CORROSION ALLEVIATION

There has been a constant attempt by engineers and scientists to improve the surface-related properties of biomaterials to reduce the failure of implants due to poor cell adhesion and leaching of ions due to wear and corrosion. The various surface modification techniques used for bioimplants have been reviewed by Anil Kurela *et al.* [90]. Preventing corrosion using inhibitors is not possible in an extremely sensitive and complex bio system and hence several coating methods have been adopted. The techniques such as chemical treatment, plasma ion implantation, plasma source ion implantation (PSII)), laser melting (LSM), laser alloying (LSA), laser nitration, ion implantation, and physical vapor deposition (PVD) and also surface texturing [28, 90, 91]. These methods are more advantageous over the other conventional techniques as they lead to better interfacial bonding, non-equilibrium phases, faster processing speed, and reduced pollution. However, each of these methods also has some limitations. Hence, some of the widely applied methods are described in the following subsections.

In the case of Ni-Ti stents, the release of nickel ions from Ni-Ti has been reported in a few cases and the released ions are found to be responsible for the endothelial cell damage. The various coating methods such as passivation, plasma immersion ion implantation, electropolishing etc used in this regard are discussed elsewhere [92-99]. Recently carbon-based coatings namely Diamond Like Carbon (DLC) are found to be more promising and the corrosion resistance of NiTi alloys with this coating has shown tremendous improvement [100].

Ti dental implants are generally surface modified to reduce corrosion, improve osseointegration and increase the biocompatibility. To achieve this, surface treatments, such as surface machining, sandblasting, acid etching, electro-polishing, anodic oxidation, plasma-spraying and biocompatible/biodegradable coatings are performed to improve the quality and quantity of the bone-implant interface of titanium-based implants [101-104]. Unlike the above treatments, laser-etching technique was introduced in material engineering originally which resulted in unique microstructures with greatly enhanced hardness, corrosion resistance, or other useful surface properties [105, 106]. Laser processing also is now being used in implant applications to produce a high degree of purity with enough roughness for good osseointegration [107]. Yue *et al.* used the excimer laser to modify the surface of the Ti-6Al-4V alloy to improve its corrosion resistance and there was a seven fold increase in the corrosion resistance [108].

With regard to orthopedic implants also, different surface modification methods have been adopted to improve their corrosion resistance [109, 110, 111-115]. Thair *et al.* studied the corrosion behavior of nitrogen ion implanted Ti-6Al-7Nb alloy by varying the dose of the nitrogen ions using an accelerator [116]. They observed that the passive current density and area of the repassivation loop were decreased as the dose values increased. Similarly the work carried out by Kamachi *et al.* on nitrogen alloying on the cold worked 316L austenitic stainless steel showed a substantial improvement in the pitting corrosion resistance [117].

Laser surface engineering (LSE) is one of the techniques employed in the area of biomaterials which is growing rapidly as it offers several advantages such as high speed, low processing time, easy to coat complex geometry, higher adhesion between substrate and the coated layer in the case of coating and in addition the surface composition can be modified without any difficulty by melting the surface in a short time. Further, laser is highly advantageous if one requires processing functionally integrated and structured materials so as to mimic the bone. The most commonly used classes of lasers are CO_2, YAG, excimer, dye, argon-ion, diode, etc. and each of these has its own unique properties and specific applications [118]. Melting of the steel and Ti surfaces using CO_2 laser is found to result in microstructures that possess high corrosion resistance and also laser nitriding of the titanium alloy in the presence of nitrogen and argon environment results in enhanced corrosion resistance [119]. Surface modification carried out on Ti-13Nb-13Zr alloy in nitrogen atmosphere by Geetha *et al*. using Nd: YAG laser in simulated body condition (Ringer's solution) was found to be significantly better for the laser nitrided samples compared to that of the untreated alloy [109]. Observation by Sathish *et al.* reveals that the corrosion resistance of laser nitrided samples is highly dependent on the processing speed as well as on the concentration of TiN dendrites and hardness [111,112].

The unique properties of nanoceramic materials have stimulated intense research so that they can be used to obtain orthopedic and dental implants with much superior properties compared to the conventional coatings which have been done hitherto with micron sized particles. Studies on corrosion behavior of nanocrystalline diamond films coated Ti-6Al-4V showed that this coating provided significant protection against electrochemical corrosion in a biological environment [113]. It has been suggested by Catledge *et al.* that this coating can enhance the life span of Co- Cr and Ti alloy implants by 40 years which is much greater than what has been achieved till now [113]. Richard *et al*. observed that corrosison resistance and fretting wear of Cp Ti increased several fold when coated with nano Al_2O_3 -TiO_2 [114]. In addition to the above, nanoceramic HAP coatings are used to enhance the osseointegration. Nanostructured graded metalloceramic coatings have also been tried to achieve better adhesion between the metal and ceramic coatings and thus nanoceramic coatings are gradually receiving greater attention.

7. CURRENT AND FUTURE DEVELOPMENT

The field of corrosion with respect to dental, orthopaedic and cardiovascular implants faces lots of challenges as there are still a number of problems to be solved. As our bone, dentin, cartilage etc are natural composites, recent research is focused on the development of composite materials for implant applications which will mimic the nature. However, more studies are to be made to understand the behavior of composite materials in *in vivo* as their biofluid absorbing behavior, interfacial bonding between the matrix and reinforcement under loading are not clear at present in *in vivo* conditions.

Ceramics are another class of materials which have high biocompatibility and enhanced corrosion resistance. They are widely used today for total hip replacement, heart valves, dental implants and restorations, bone fillers and scaffolds for tissue engineering, but ceramics are brittle, have high elastic modulus and can fracture as they posses low plasticity. In addition, when they are oxidized they release ions into the body and this may lead to degradation of the implant [120]. Alumina and zirconia are considered to be as alternatives for metallic materials for load bearing applications as they show no corrosion in the body and also posses high wear resistance. But mechanical failures of these implants are also being reported and hence extensive research is needed in order that these classes of materials are being recommended for the final applications. Bioactive glass which was first discovered in 1969 is now used widely for bone repair and bone regeneration. Fused quartz, aluminosilicates, certain borosilicate, alkali resistant glass, soda-lime glass, titania frit, arsenic trisulfide, lithium and magnesium aluminosilicate, glass-ceramics, and calcium-fluorapatite all appear to be well tolerated and seem acceptable for soft tissue implantation. Though they exhibit very low corrosion a more sensitive method to study low corrosion rates of glasses is yet to be devised [121].

Several interactions mentioned earlier which lead to biological corrosion should be understood at the atomic level. Though there are standards available to test the corrosion performance of the materials being developed there are always variation in the methodology adopted by different research groups. Uniform methodology should be adopted to compare the results of the different groups working in this direction. As the *in vitro* test can be considered only as screening test as it could not give the real picture, a simulator has to be developed with all facilities to measure corrosion in the simulated body condition like hip and knee simulators which are often used to test only the tribological properties of the materials. The current simulators employed should include testing facilities to measure tribological corrosion also to have the actual picture of the various processes taking place in real time. Though there are many reports which show the adverse effects of the corroded products, still, there is a need to develop a methodology to evaluate the actual concentrations, the form of the metals that will induce toxicity and other adverse effects.

Surface modifications are often performed on the biomedical implants to improve corrosion resistance, wear resistance, surface texture and biocompatibility [122, 123, 124]. All the modified surfaces should be tested for its corrosion behavior invariably apart from improving other desired properties. A thorough understanding of the interactions which take place at the atomic level between the surface of the implant, the host and the biological environment including all types of micromotions of the implants kept inside the human system should be studied carefully in a greater detail in order to obtain implants which can sustain for a longer period in the human system. In final, one has to admit that so many owe to few who have toiled tirelessly and succeeded in relieving them from their sufferings and to increase their longevity and man's mission to conquer the unconquerable will continue forever in spite of the known fact that it is impossible by these replacements that one talks about, can never replace or reproduce nature. The field of corrosion in biological systems is young and fertile as man knows only little about his physiology and its interactions with the foreign body is much more complicated and hence the mission will continue.

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