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ORIGINAL PAPER

Wear and Corrosion Interactions on Titanium in Oral Environment: Literature Review

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Abstract The oral cavity is a complex environment where corrosive substances from dietary, human saliva, and oral biofilms may accumulate in retentive areas of dental implant systems and prostheses promoting corrosion at their surfaces. Additionally, during mastication, micro-movements may occur between prosthetic joints causing a relative motion between contacting surfaces, leading to wear. Both processes (wear and corrosion) result in a bio-tribocorrosion system once that occurs in contact with biological tissues and fluids. This review paper is focused on the aspects related to the corrosion and wear behavior of titanium-based structures in the oral environment. Furthermore, the clinical relevance of the oral environment is focused on the harmful effect that acidic substances and biofilms, formed in human saliva, may have on titanium surfaces. In fact, a progressive degradation of titanium by wear and corrosion (tribocorrosion) mechanisms can take place affecting the performance of titanium-based implant and prostheses. Also, the formation of wear debris and metallic ions due to the tribocorrosion phenomena can become toxic for human tissues. This review gathers knowledge from areas like materials sciences, microbiology, and dentistry contributing to a better understanding of bio-tribocorrosion processes in the oral environment.

Keywords Titanium · Bio-tribocorrosion · Wear · Corrosion · Dental implants

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1 Introduction

Several machines, installations, and devices used in different areas can be damaged by wear and corrosion processes [1, 2]. Medical devices and apparatus are also vulnerable to wear and corrosion phenomena, as reported in literature [3–6]. Most often, there is a loss of material that can lead to the deterioration of the performance of that device, machine, or installation [1]. Artificial organs, including dental implants and prostheses, possess a major drawback of a limited lifespan due to friction, wear, or decay of structural materials in the warm, humid, and corrosive environment of the human body. The simultaneous degradation of biomaterials by wear and corrosion in the oral cavity is a phenomenon that has been studied in order to prevent failures of dental restorations, implant, and prostheses, to avoid eventual detrimental effects to the

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patients, and to improve the function of oral rehabilitation systems. Therefore, the study of wear and corrosion resistance of structural materials can determine the performance of dental implants and prostheses. As a result, the reduction of restorative material loss by degradation can increase the long-term success of dental implant systems.

Tribocorrosion is a term used to describe the irreversible transformation of a material caused by a simultaneous action of chemical, mechanical (wear), and electrochemical (corrosion) interactions on surfaces subjected to a relative contact movement [1, 2, 7, 8]. Thus, a corrosive environment can amplify the material loss rate by wear mechanisms as well as, inversely, wear can increase the corrosion rate [1, 9]. The tribocorrosion rate of metallic biomaterials such as titanium depends on the mechanical and chemical properties of their oxide film as well as on the environment [1, 9, 10]. Nonetheless, titanium oxide passive film can be destroyed by bending or wear mechanisms (fatigue, abrasion, adhesive wear, fretting) exposing the underlying metal [11]. However, tribocorrosion can be beneficial in manufacturing technology like grinding and chemical–mechanical polishing of biomaterial surfaces in the fabrication of dental restorations, implants, and prostheses.

Nowadays, the tribocorrosion behavior of materials has been studied in biological environments originating the new designation of bio-tribocorrosion [3, 12–15]. Even though it is not possible to simulate the complex oral environment for bio-tribocorrosion tests, in vitro studies can, at least, determine the influence of each component on bio-tribocorrosion behavior of biomaterials. For instance, Guindy et al. [16] reported the failure of six dental implant systems caused by corrosion of the prosthetic superstructure. In that study, areas with clear signs of localized corrosion were detected by scanning electron microscopy on all implants and inner crown surfaces [16]. Another previous study reported a synergistic effect of distinct mechanisms, which led to total failure of implants under extrinsic common fatigue loading [6, 17]. The propagation of a crack from the body of the retrieved implant was detected by scanning electron microscopy [6].

The tribocorrosion behavior of materials is influenced by several aspects related to contacting materials, mechanics of the tribological contact, and physico-chemical properties of the environment [1, 2]. Mechanical aspects such as applied forces, contact geometry, and type (sliding, fretting, rolling, or impact) determine the tribocorrosion rate for a given material [1–5]. Concerning contacting surfaces, the topography (e.g., roughness, adsorbed molecules, and oxide film properties), chemical composition, and microstructure (e.g., phase distribution, grain size, etc.) of materials play an important role in the tribocorrosion system [1–5]. Finally, the corrosion of a material depends on the chemical composition, pH, temperature, and presence of oxidative species in a gaseous or liquid environment [1].

2 Structural Materials for Dental Prostheses and Implants

It is important to mention that different structural materials can be used in dental implants and prostheses. For instance, commercially pure (CP) titanium is frequently used to fabricate dental implant fixtures (Fig. 1). The clinical long-term success of dental implants is related to their osseointegration [18, 19]. Different methods are used to modify the titanium surfaces accelerating the osseointegration of dental implant fixture surfaces, such as grit-blasting, acid-etching, anodization, or calcium phosphate coatings [18].

The abutment can be produced from titanium-based alloys or else from ceramic materials such as yttria partially stabilized tetragonal zirconia polycrystalline (YTZP). Abutments made of YTZP provide better results concerning esthetic for anterior restorations although data on long-term performance of fixed partial dentures on YTZP are still lacking [20, 21].

Additionally, ceramic materials are used to produce metal-free (feldspar-based porcelain fused on YTZP infrastructure) and metal-ceramic (e.g., feldspar-based ceramic fused on metallic infrastructure) crowns (Fig. 1) [22]. For a given material, a variation of properties may exist as shown in Table 1.

Such variation of properties can be explained by differences in microstructure and/or residual elements. A match of mechanical properties between materials used in implant systems is fundamental in oral rehabilitation. For instance, the wear rate of structural materials can be higher when there is a large difference in hardness between abutment and implant fixture or between abutment and crown joints. Thus, the relative importance of mechanical, physical, or chemical properties will depend on the biomaterial application. A dental implant-supported prosthesis should possess mechanical properties close to that of dental and bone structures in order to establish a long-term clinical performance and harmony with the masticatory system.

Nowadays, different dental implant-abutment joints are commercially available for implant-supported prostheses

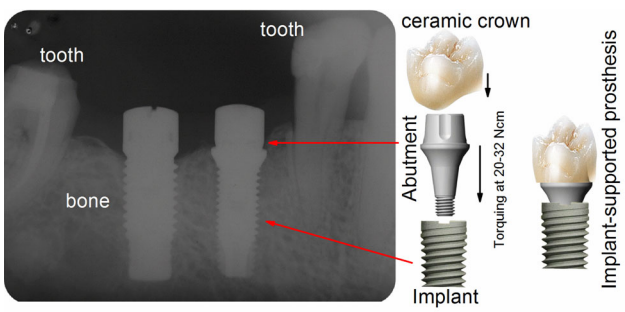

Fig. 1 Schematics of a titanium-based abutment-implant joint

Table 1 Mechanical properties of materials used in oral rehabilitation compared to those of natural materials: enamel, dentin, and cortical bone [33–38, 122]

Materials	Tensile strength (MPa)	Elastic modulus (GPa)	Vickers hardness (HV)
CP Ti grade 2 (α-titanium)	345	102–119	180–209
Ti–6Al–4 V (α + β–titanium)	895–930	110–150	350
Ti–15Zr–4Nb–4Ta–0.2Pd (α + β–titanium)	715–919	94–99	250–350
CoCrMo	560–690	180–240	317–460
Gold alloy (type IV)	410–770	95–123	235–360
Dental porcelain	34–82	66–82	443–780
Enamel	10	75–100	300–410
Dentin	52	18–19	80–92
Cortical bone	140	10–18	43–76

such as hexagon (external or internal) or Morse taper connections. In an implant-supported prosthesis, an excellent fit between titanium-based abutment and implant (Fig. 1) results in a proper mechanical integrity and distribution of masticatory forces [23, 24]. The fit of the implant-abutment assembly is dependent on machining process and properties of the structural materials.

On the other hand, the poor fit of dental implant-based joints can result in a higher displacement of the structural parts under mastication forces or occlusal prematurity from incomplete seating [23, 24]. Binon and McHugh [23] reported on the loosening of abutment screw joint due to the implant-abutment rotational hexagonal misfit. Failures in dental implant systems have been attributed not only to biomechanical overloads but also to corrosion and wear synergy along with the cyclic loading mechanism of the masticatory process [6]. Nevertheless, it is difficult to correlate failures of dental implant systems with bio-tribocorrosion mechanisms in vivo. Therefore, corrosive substances can accumulate in the internal connection of dental implant systems and also in the biofilms formed on external and inner surfaces of the prosthetic gaps [25, 26]. The pH lowering associated to corrosive substances and under mechanical solicitations can decrease the long-term performance of dental implant systems [6, 13, 16]. In addition, polished surfaces can become rough in the oral cavity due to the effect of food debris or due to the friction between contacting surfaces increasing biofilm accumulation [9, 12, 13].

3 Mastication Forces and Distribution of Stresses Through Structural Materials

Mastication forces produced during the chewing cycle have been described to be in the range of 10–120 N [27, 28]. Nevertheless, the properties of the food bolus (thickness, elastic modulus, hardness) as well as human body features (muscle activity, gender, age, weight, presence of other dental prostheses) influence the magnitude of mastication forces generated on dental surfaces [27–31]. The highest mastication forces are generated at the end of the chewing cycle when sliding motion stops as the teeth reach the centric occlusion that produces localized abrasion wear of contacting dental surfaces [27, 28]. In literature, the maximum biting forces were measured by different methods (e.g., electromyography, occlusal transducers) and are in the range of 89–150 N at the incisors (anterior region), 133–334 N at the canines, 220–445 N at the premolars (intermediary region), and 400–600 N at the molars (posterior region) [32–34].

A dental implant-supported prosthesis must present a noteworthy ability to sustain mastication forces that depend on the design and structural materials properties. The orientation of stresses is very important once axial loads promote the transfer of stress through dental implant systems to the bone tissue [19, 21]. However, oblique loads can originate overload on structural materials and on bone tissue that can promote failures by fatigue and wear of the implant-based system [6, 24].

In addition, the presence of different materials provides abrupt variations of different properties (hardness, elastic modulus, yield strength). Also, aspects related to the design of the implants such as length, diameter, and shape can be adjusted to decrease the stress distribution to the bone [19, 36].

Computer simulations have been developed to allow evaluation of the loads distribution through dental implants systems and prostheses that could lead to material and peri-implant bone loss [35–38]. Papavasiliou et al. [35] revealed, by three-dimensional finite element analysis of stress distribution around single tooth implants, that the highest stresses were concentrated in the cortical bone [35]. On axial and oblique loading at 20 N, the highest stresses in the bone (12–16 MPa) were below the elastic limit of cortical bone (about 60 MPa). However, on loading at 200 N, resolved stresses on the cortical bone were higher than that elastic limit of bone. Also, high stress values were found at the implant-abutment joint (Fig. 1) in the range of

9 up to 18 MPa, and 110 up to 170 MPa on oblique loading at 20 and 200 N, respectively. However, the values were lower for axial loading at 20 N (0.5–0.9 MPa) and 200 N (5–9 MPa) [35]. Applying 100 N static axial occlusal loads, Eraslan and Inan [37] also noticed a high concentration of von Mises stresses located at loading areas of abutments and cortical bone for all models [35]. Baggi et al. [36] found numerically the highest von Mises stress values (ranging from 65 to 220 MPa on vertical loading at 250 N) at the titanium implant crest level (area between abutment and bone) that decreased for implants with large diameters [36]. Alkan et al. [38] found von Mises stress (on oblique loadings at 70 N) at titanium abutment screws in the range of 80 up to 145 MPa.

As there is no periodontal ligament around implants such as in natural teeth, the shock-absorbing ability of dental implants is lower than that of natural teeth [33]. Thus, an intra-mobile element (screw thread) of titanium is often included to decrease the stress distribution to the bone [19, 35] although micromovements take place in the prosthetic joints [24].

4 Electrolyte in the Oral Cavity: Human Saliva

Human saliva consists of a mixture of fluids produced from parotid, submaxillary, and submandibular glands as well as by oral mucosal glands (labial, lingual, palatal, and vestibular glands) at a pH between 6 and 7 [39, 40]. The composition of saliva, which includes organic, inorganic compounds, and 99 % water, is also dependent on external factors that can be present in the oral cavity [39, 41]. Surfaces inside oral cavities are regularly reached by saliva at a pH altered, between 3 and 8, by external factors such as dietary, presence of acidic substances, and microbial metabolites [39, 41]. Additionally, the composition and properties of saliva can be modified by internal factors associated to salivary gland dysfunctions or to the time of the day [39, 41]. The role of saliva has been considered in the maintenance of the oral health of the human body due to the presence of numerous organic and inorganic compounds [39, 41]. Proteins (e.g., albumin, proline-rich proteins, statherin, histatin), glycoproteins (e.g., mucin), and aminoacids (e.g., leucine, glycine, glutamate, aspartate) are the main organic constituents of the saliva and valuable for microorganisms [39, 41, 42]. Even though some organic constituents are important for microbial metabolism and growth, other constituents such as antibodies (IgAs, IgM, IgG) and enzymes (lyzozyme, lactoferrin, lactoperoxidase) act as regulators of microbial colonization. Additionally, carbohydrates (glucose, galactosis, and sialic acid) and lipids (phospholipids, triglycerides, and cholesterol) are also organic constituents present in the saliva.

In the oral cavity, the viscous property of the saliva provided by glycoproteins (e.g., mucin) present in the acquired pellicle can protect the dental surfaces against wear [12, 39]. The friction recorded on titanium surfaces under sliding against an alumina ball can be reduced in the presence of water, lipids, and glycoproteins (e.g., mucin). That can be compared to the effect of commercial lubricant agents [12]. However, there are few studies on the bio-tribocorrosion in simulated oral environments containing glycoproteins, such as mucin and albumin both present in saliva [3, 12, 13, 15, 43, 44].

Previous studies have reported the bio-tribocorrosion behavior of titanium alloys in a buffered solution containing lypopolysaccharides (LPS) or albumin [15, 43, 44]. Khan et al. [44] revealed that the wear rate of Ti13Nb13Zr, Ti6Al7Nb, and Ti6Al4V decreased in the presence of albumin. Contrarily, Hiromoto and Mischler [43] did not find any effect of albumin on the fretting-corrosion behavior of titanium [43]. Mathew et al. [45] reported a negative effect of LPS on the corrosion/wear behavior of titanium in artificial saliva. Additionally, the presence of LPS can induce an accumulation of biofilms.

The inorganic fraction is basically represented by ions such as Ca^{2+}, PO_4^{3-}, Na^+, K^+, and HCO_3^-. Bicarbonate (HCO_3^-) and phosphate (PO_4^{3-}) ions act as a buffer to maintain the pH of the saliva between 6 and 7 [39, 46]. Acting as the main buffering agent, HCO_3^- binds to H^+ to form H_2CO_3, H_2O, and CO_2, increasing the pH which leads to the prevention of tooth demineralization and corrosion of dental materials [39, 41, 46]. However, the buffering mechanism can be limited by a high density of microbial cells or by a low salivary flow rate [39, 41, 46].

The salivary glands produce 1–1.5 L of saliva per day which is responsible for the mechanical removal of microorganisms and food stuffs. The masticatory process and the muscular movements increase the salivary output optimizing the oral cleaning [39–41, 46]. However, the salivary flow rate decreases during sleep facilitating the increase in the number of microorganisms in the oral cavity and consequently the lowering of the pH [39, 41]. In fact, the increase of lactic acid-producing bacteria metabolism is a critical factor for the lowering of the pH.

In order to mimic human saliva, several artificial saliva solutions have been used to study the corrosion behavior of dental materials [47–49]. A previous review of nearly 60 artificial saliva recipes was carried out to clarify the role of the compounds used in the artificial saliva formulation [48]. That previous study focused on the buffer effect, the role played by CO_2 gas and the presence of calcium ions, hydrogenocarbonates, hydrogenophosphates, and thiocyanates. The pH of the artificial saliva solutions found in literature ranged from 4.5 to 7 [47, 48]. Due to the inconsistent and unstable properties of natural saliva, the

formulation of artificial saliva solutions that react with the test material in a way similar to that of natural saliva is not easy to achieve in vitro [47, 49]. Most of the reported artificial saliva solutions are a simplified version of what may actually occur in the oral cavity in terms of solubility of components and corrosion of dental materials [47, 48].

The use of organic-free artificial saliva solutions has been often applied in corrosion studies [47–49] and tribocorrosion studies [9–13, 50, 51]. Concerning the corrosion behavior of dental materials, the properties of several artificial saliva solutions were reported in literature [47–49, 51, 52]. However, some previous studies [51, 52] reported that the corrosion behavior of dental materials in artificial saliva proposed by Fusayama [53] was most closely approximating the one in natural saliva. Nevertheless, other formulations have also been reported as most appropriate for studying corrosion of dental materials [49]. Even though the extensive number of different artificial saliva formulations is found in the literature, Fusayama's solution has been largely used to study the corrosion of dental materials including titanium and its alloys [49, 50, 54–57]. Concerning several parameters such as the use of artificial saliva, studies should be standardized in order to compare the results reported in the literature.

5 Presence of Oral Biofilms

Previous studies have revealed the effect of biofilm on the tribocorrosion in the oral cavity concerning the following factors: (1) biofilm composition, (2) biofilm adhesion process, (3) role of microorganisms, (4) biofilm metabolites, and (5) restorative surface characteristics. The oral cavity is a complex environment that gathers several substances from food and saliva to microorganisms and their metabolites [39]. Along time, several areas in the oral cavity can be covered by a complex microbial community embedded in an extracellular matrix composed of polysaccharides, proteins, nucleic acids, and water, known as oral biofilm [58–60]. As a result, the pH in the oral cavity is frequently altered reaching low values after the intake of acidic substances and/or acids release from oral microbial metabolism [13, 39, 50]. Moreover, the biofilm composition is influenced by the local pH values, considering the release and tolerance of bacteria to acids [39, 41, 50, 61]. The temperature also varies temporarily during the intake of warm or cold foods. Therefore, there is a variation of oxygen in the oral cavity, as for example the low presence or absence of oxygen concentration in the areas below gingival margin. As a consequence, the microbial colonization in the mouth follows the variation of oxygen which promotes the preferential growth of aerobic or anaerobic microorganisms [39, 41, 62, 63]. Finally, the oral cavity habitat must not be considered as uniform since there are different micro-areas depending on the saliva composition, nutrient accumulation, tissue and restorative surfaces, and resident microorganisms [39]. The topography of dental restorative systems is of major importance for microbial colonization taking into account that rough surfaces are more susceptible to be colonized by microorganisms than smooth ones [64–68].

In the oral cavity, microbial adhesion can take place in both soft tissues and hard structures represented by tooth and restorative structures. These surfaces are usually coated with a conditioning film (0.1–10 μm) (Fig. 2a) that

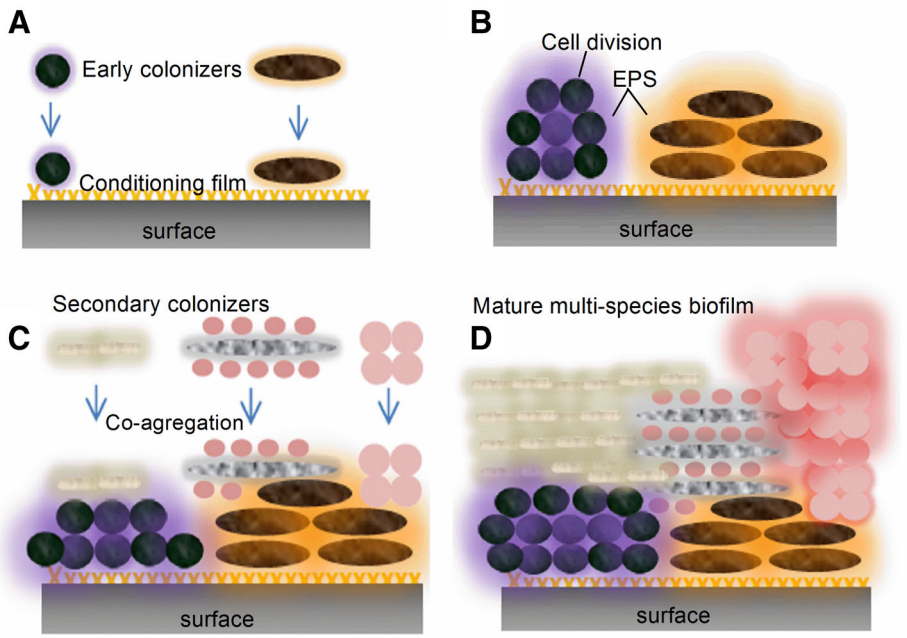

Fig. 2 Schematic biofilm formation and co-aggregation of multi-species biofilms (**a**) Initial biofilm formation by primary colonizers on a substratum covered with a conditioning film; **b** cell growth, division, and production of extracellular matrix (EPS); **c** coadhesion of single cells; and **d** maturation and the formation of the multi-species biofilms

is composed of glycoproteins, ions (e.g., Ca^{2+}, Mg^{2+}), and water [39, 41, 69]. The conditioning film or enamel acquired pellicle, such as often known when covering tooth enamel, protects the oral surfaces against wear originated from masticatory contacts and determines the adherence of microorganisms [39, 41, 61, 67]. However, the primary microorganism colonizers present protein macromolecules on their surfaces named adhesins that bind to receptors present on glycoproteins (e.g., mucin) in the conditioning film at oral surfaces [39, 41, 61, 67, 70]. This is a specific mechanism of microbial colonization that allows microbial cells to bind selectively to surfaces (Fig. 2) [61, 69, 70].

Steptococcus species such as *S. sanguinis*, *S. oralis*, *S. gordonii*, *S. mitis*, *S. mutans*, and *S. sobrinus* represent 60–80 % of all primary colonizers, which also include 5–30 % species of *Actinomyces naeslundii*, *Fusobacterium nucleatum*, *Capnocytophaga ochracea*. Different adhesins are present in the adherence of *Streptococcus* species and acquired pellicle. *S. sanguinis* and *S. oralis* possess adhesins similar to lectine cellular membranes, which are called lectins. For instance, *S. gordonii* presents more than one adhesin that binds at least to three receptors, namely proline-rich proteins, salivary agglutinins, salivary amylase [39]. In order to colonize host oral surfaces, *Streptococcus* can use different mechanisms. At a first stage, *Streptococcus mutans* establishes electrostatic interactions with salivary glycoproteins receptors mediated by Ca^{2+} [39, 41, 70]. Additionally, it can occur a binding between and glycoproteins present on mucin which is part of the acquired pellicle [42, 68, 70, 71]. Moreover, these bacteria are able to produce hydrated extracellular polysaccharides (EPS), resulting from sucrose degradation by enzymes known as glycosyltransferase (GTF), as shown in Fig. 2b [39, 41]. EPS is composed of polysaccharides chains α-1,3 and α-1,6 glucan linkages that bind to receptors of *S. mutans* represented by GTFs, and promote the agglutination of *S. mutans* cells [39, 72]. The proteoglycans and signaling molecules control the homeostatic dynamic state of the entire extracellular matrix [58].

The multilayered biomass composed of glycoproteins, water, nucleic acids, and polysaccharides chains acts as viscoelastic material which can support considerable elastic deformation under shear stresses and is able to distribute loads, thereby decreasing the contact pressure at the surface [12, 13, 73–75]. Also, that viscoelastic biomass can be responsible for the friction on oral surfaces [12].

Other microorganisms such as *S. sanguinis*, *S. gordonii,* and *S. oralis* produce EPS composed of glucans although there is a lower agglutination in these cells than in the *S. mutans* biofilm [39, 41]. Furthermore, the glycoproteins present in saliva and gingival fluid can support the co-aggregation between different species like between *C. albicans* and *S. mutans* or among *S. sanguis*, *S. oralis,* and *A. naeslundii* [39, 41]. Also, the cell–cell co-aggregation can occur by adhesin–receptor interactions [70]. Since there is a modification of the environment associated to the presence of early colonizers, secondary or late colonizers can co-aggregate with previous species forming multi-species biofilms [61, 63] as shown in Fig. 2c, d. For instance, late pathogenic colonizers such as *Prevotella intermedia* and *Porphyromonas gingivalis* can co-aggregate with filamentous (*Actinomyces naeslundii*) and fusiform (*Fusobacterium nucleatum*) bacteria that can bind to glycoproteins in the acquired pellicle or to other primary colonizers [39, 41, 61, 63]. Finally, the cell growth and division in a complex microbial community follows nutritional and environmental conditions in the oral cavity [61, 63].

Leonhardt et al. [76] evaluated the early bacterial colonization on titanium, amalgam, and hydroxyapatite in vivo, and no significant quantitative and qualitative differences in bacterial colonization of these materials were found. However, Rosentritt et al. [77] reported significant differences between the *S. mutans* colonization on ceramic, composites, and alloys in vitro. These authors described that adhesion was higher on composites than on alloys which corroborates the results of Tanner et al. [71].

The microbiota present at peri-implant seems to depend on the same factors related to microbiota of natural tooth surfaces [78–83]. The highest concentration of microorganisms (70 %) is represented by Gram-positive coccus and facultative anaerobic bacillus [78]. Therefore, the commensal microbiota present in the oral cavity influences the microbial colonization of dental implant systems and prostheses [41]. For instance, Mombelli et al. [84] reported the presence of pathogens such as *P. gingivalis*, *P. intermedia*, and *Fusobacterium* in peri-implant microbiota of partially edentulous patients with a history of previous periodontal disease [84]. On the contrary, Danser et al. [85] did not find *P. gingivalis* in peri-implant areas of 30 edentulous patients with a history of periodontal disease. Analyzing subgingival areas of 18 unsuccessful implant systems, Alcoforado et al. [86] detected the presence of *C. albicans* on 5 implant systems, while *P. intermedia* was found only on 4 implants [86]. Rosenberg et al. [87] reported the presence of *C. albicans* in 10 % of peri-implant microbiota that also comprise *P. gingivalis*, *P. intermedia*, and *Fusobaterium* [87]. Leonhardt et al. [76] also found the presence *C. albicans* in microbiota associated to peri-implant inflammations. These findings seem to correlate the incidence of opportunistic infections by *C. albicans* due to the use of antibiotics for peri-implant infections before the removal of implant systems [41].

In a dental implant-supported fixed prosthesis, the microbial colonization begins at prosthetic areas exposed to the oral environment taking into account that biofilm formation depends on the prosthetic design, surface

conditions, and on the oral microbiota [41, 66, 67, 88]. After implantation, a part of the margin area of implant fixture is in contact with connective and epithelial tissues, while another part is in contact with abutment and oral fluids. In literature, a mean interfacial discrepancy of about 2.5–60 μm in implant fixture-abutment gaps was reported [25, 26, 89–92]. As the diameter of microorganisms is less than 10 μm, the prosthetic gaps can be effortlessly colonized by several microorganism. Hence, the penetration of microorganisms in implant internal connections can be caused by microbial leakage at the implant-abutment joints [26, 92].

Previous studies reported instability by unscrewing of about 50 % of abutment screws analyzed for 1 year [93]. The presence of oral fluids and biofilms in the implant internal connection and prosthetic microgaps [12, 25, 90, 94] can be one of the factors responsible for a loss of mechanical integrity of the abutment screw by unscrewing. Thus, biofilms generated an ultra-low friction on titanium under sliding [12]. On the other hand, as a result of biofilm growth, there is a release of acidic substances from carbohydrates metabolism that alters pH and the oxygen content of the local environment [11]. Specifically, lactic acid-producing bacteria such as *S. mutans* perform fermentation of carbohydrates (e.g., sucrose) releasing lactic acid that decreases the pH to values lower than 5.5 and dissolves the carbonate hydroxyapatite mineral of teeth by a process called demineralization. It was also reported that *S. mutans* can promote lowering of pH to 4.0, while *S. mitis* and some species of *Lactobacillus* promote lowering of pH to 4–5 and 3.0, respectively [41, 50, 93, 95]. However, the pH of the oral surfaces surrounding media can be lower than the ones reported that could promote a localized corrosion of titanium.

The localized corrosion of titanium caused by biofilm colonization has been revealed by previous studies [16, 56]. Mabilleau et al. [56] reported a localized corrosion of titanium in vitro after 21-day immersion in a medium containing *S. mitis* cells. Also, Souza et al. [13, 50] reported the decrease of corrosion resistance of titanium in the presence of *S. mutans* or *S. mutans/C. albicans* biofilms by electrochemical tests [13, 50]. In fact, the pH of the medium in which biofilms grow decreased in the presence of microorganisms probably due to the release of acidic substances that reduced the corrosion resistance of Ti [13]. The exposure of structural materials to oral fluids, including acidic substances produced by bacterial metabolism, is associated to the corrosion of the implant fixture-abutment joint [16]. In addition, fluorides can be accumulated in biofilms depending on their structure and composition, physico-chemical properties of the solute, and biofilm thickness [96–99]. Due to the diffusion of F^- ions through extracellular matrix, fluorides can also reach oral tissues and other micro-areas in the biofilm [96, 97]. The lowering of pH caused by the release of lactic acid from microbial metabolism in the biofilm can be responsible for a considerable concentration of HF that can corrode titanium and feldspar-based porcelain surfaces of dental implant-supported prostheses.

6 Corrosion of Titanium

Since the intensive work accomplished by Branemark et al. [100], titanium and its alloys have been the first-choice materials for implant systems (implant fixture, abutments) and prostheses (dental metal-ceramic crown and removable denture frameworks) in oral rehabilitation [19–21, 100].

Titanium is known as a material with a very high corrosion resistance in physiological solutions, and has an excellent biocompatibility due to the formation of a protective titanium oxide film, like TiO_2, when in contact with the surrounding environment [10, 18, 100, 101]. Also, properties such as low density (4.5 g/cm^3) combined with low thermal–electrical conductivity and high mechanical strength are often referred in literature and thus uphold titanium alloys as a material remarkably required in medicine and dentistry. The passive TiO_2 formed on titanium surfaces is found to be amorphous. However, rutile or anatase on TiO_2 films can be created, for instances, by heat treatments. Both titanium dioxide films present a tetragonal form and perform an important role upon corrosion and biocompatibility [101, 102]. The TiO_2 film possesses a high corrosion resistance in various test solutions, such as artificial saliva, Ringer's solution, 0.9 % NaCl solution, or physiological saline solution [54, 55]. Nevertheless, the protective TiO_2 film can degrade in the oral cavity in the presence of corrosive substances such as fluorides, lactic acid, carbamide peroxide (urea peroxide), and hydrogen peroxide [10, 54–56, 103]. The breakdown of the titanium passive film leads to a localized corrosion failure such as intergranular attack, pitting, or corrosion fatigue [101, 104].

The degradation of titanium-based surfaces at high fluoride concentrations was found in previous studies revealing the occurrence of a localized corrosion process, namely pitting corrosion [10, 55, 56]. The occurrence of pitting corrosion was described as resulting from the formation of hydrated Ti oxides as $Ti(OH)_2F^+$, and salts as $[TiF_6]^{2-}$, TiH_2, $Na_3Ti_3F_{14}$, TiF_4 $[TiF_6]^{3-}$ in the presence of HF [10, 55]. Such previous studies have revealed that a minimum concentration of 30 ppm HF is enough to promote a localized corrosion of titanium in fluoride solutions. In fact, the corrosion in fluoride solutions depends on the pH and the formation of HF produced by the dissociation of NaF when it is present at high concentrations, or in low pH solutions due to the bonding between H^+ and F^- ions. For instance, a localized corrosion on titanium surfaces might occur in a solution containing 452.5 ppm F^- at pH 4.2 or in a solution containing 227 ppm

F^- at pH 3.8. The formation of pits on CP titanium was also found in a previous study after immersing it in artificial saliva containing 11,180 [56] or 12,300 ppm F^- [10]. Chemical analyses indicated the presence of heterogenic oxides as $TiOF_2$ and TiOHF on CP titanium after immersing it at high F^- concentration [10].

Considering the titanium surface modification, chemical methods are useful to increase the surface roughness of dental implant fixtures in order to promote osseointegration. Such methods include etching of surfaces using acidic substances such as HCl, H_2SO_4, HNO_3, and HF. Micro- or nano-porous surfaces can also be produced by potentiostatic or galvanostatic anodization of titanium in acidic substances (H_2SO_4, HNO_3, H_3PO_4, and HF) at high current density or potential application, resulting in a thick and porous titanium oxide layer ($\sim$1 μm) depending on the parameters. Also, anodizing in an electrolyte-containing phosphate and calcium-based solutions can promote the formation of thick and porous layer composed of calcium and phosphates [18].

7 Wear Processes on Dental Materials

During chewing process, abrasion of restorative surfaces including titanium can be caused by frictional surface interactions with opposing surfaces, toothbrush and paste, food bolus, and hard particles originated from dietary [30]. A two-body abrasion has been reported when two surfaces were rubbed away from each other by direct contact with their asperities [92, 105, 106]. In the oral cavity, two-body abrasion takes place during a "non-masticatory tooth movement" [30] although it can occur in the prosthetic joint surface during masticatory tooth movement. Moreover, the presence of "intervening slurry of abrasive particles" in the tribological contact originates in the three-body abrasion [105, 106]. Under high or low stresses, this kind of mechanism occurs during the masticatory process due to the presence of abrasive particles in the food bolus [30, 31] or it can occur during the wear process of dental surfaces with material loss and debris formation [9, 105, 106]. Then, abrasive particles move along the surfaces in tribological contacts scratching away the antagonist surface [9, 105, 106]. If the prosthetic joints act as a closed tribological system, the material loss will be higher than the loss in open systems where the abrasive particles move away from the tribological contact [9].

The wear phenomenon known as fatigue consists in a rupture of intermolecular bonds and a zone of subsurface damage caused by the movement of surface molecules under cyclic loads [9, 105, 106]. Consequently, there is a micro-crack formation within the subsurface oblique to the surface, which can coalesce to the surface, and thus material loss can occur [106]. Fatigue has been often associated to wear of occlusal surfaces [105]. Another wear mechanism, known as adhesive wear, occurs when, after oxide film disruption, promoting an attraction between two surfaces that are under relative contact motion. Wear particles can also be attached like platelet shapes to surfaces under friction. However, fractures of the micro-welds resulting from adhesive wear can occur and can increase the wear rate [105, 106]. Fretting is also an important wear mechanism that can occur between contacting surfaces under small-amplitude oscillatory movement [5, 43, 105, 106, 113]. The movement can result from one of the contacting members undergoing cyclic stress, and it can reduce the fatigue strength by 70–80 % [106]. Fretting wear has been associated to wear of cortical bone against titanium implant surfaces [30].

8 Simultaneous Degradation of Titanium by Corrosion and Wear Interactions

Friction on titanium during mastication can destroy the TiO_2 film that leads to a material loss [1, 9, 13, 107] and possible failures of dental implants and prostheses [6, 17]. As a result from corrosion and wear processes, metallic ions are released, and wear particles originating from titanium were found in the surrounding tissues and associated to inflammatory reactions [106, 108–111].

Different tribocorrosion mechanisms can take place during rubbing between a ductile metal (e.g., titanium) and a hard inert counter-body (alumina), as shown in Fig. 3 [1].

Mechanical and electrochemical mechanisms are responsible for the material removal from the hard less materials (first body) during rubbing [1]. As a result, there is plastic flow with metal ejection by plowing and metal detachment forming third bodies (wear particles) [7, 8, 107]. The wear particles can be transferred and deposited on the alumina surface or spreading on the titanium surface by adhesive wear forming tribolayers [1, 9, 107]. In contact with environment, the wear particles can be oxidized and

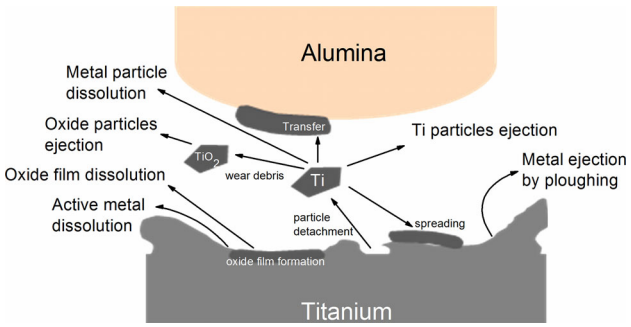

Fig. 3 Schematic tribocorrosion mechanisms of titanium. Adapted from Landolt [1]

form solid oxide that can modify the mechanic of contact. Then, a brittle oxide particle can be formed contributing to a third-body abrasive mechanism and can extend the mechanical wear of titanium. On the other hand, a solid oxide can chemically dissolve as ions in the environment taking into account that there is also titanium dissolution and an ion release produced by electrochemical reactions between titanium and the environment [1]. The metal detachment exposes a fresh titanium surface that reacts immediately with the environment corresponding to an anodic partial current and a subsequent increase of the corrosion rate due to the high chemical reactivity of bare metal [1, 2, 9, 93]. Then, a galvanic cell is established during the wear process in the electrolyte, where the bare metal (worn area) may act as an anode or a cathode, and its periphery, represented by the passive layer (unworn area), acts as a cathode or an anode, respectively [2, 9, 107, 112]. Consequently, there is a current flowing between anodic and cathodic areas, which induces an electrochemical potential distribution over the surface [2, 112].

In fact, the chemical and mechanical properties of the titanium passive film influence the surface mechanical response of titanium as well as the third-body behavior [8–10, 50]. This comprises the repassivation rate of titanium that consists in the formation of a new TiO_2 film immediately after its mechanical destruction (depassivation) [8, 9]. Barril et al. [113] studied the fretting corrosion of Ti6Al4V in 0.9 % NaCl solution and revealed a strong influence of the electrode potential on the wear rate of titanium alloys. In addition, it was revealed that the oxidation of third-body particles at anodic potentials decreases the mechanical energy involved in the wear process. Considering the presence of fluorides, the wear processes on titanium in high fluoride solutions (12,300 ppm F^-) are quite different compared to the ones noticed in artificial saliva without or containing up to 227 ppm F^-. In fact, the formation of a reaction product layer on titanium at high F^- concentration decreases the coefficient of friction [9]. However, a progressive corrosion of titanium has been detected by surface analysis, as well as by electrochemical measurements, indicating an active state of titanium in artificial saliva at high F^- concentration. The wear rate of titanium in sliding contacts was too fast at high fluoride concentration which could occur in titanium-based structures used in prostheses and dental implants. This last case could be a cause for failures of titanium-based implant systems considering that the material loss can increase microgaps in the prosthetic joints and modify the contact area of structural materials. As a consequence, the distribution of loads on the implants could be altered promoting over-loads at certain contact areas. Additionally, over-loads can increase the wear rate of prosthetic materials exposed to relative contact motions.

In the presence of biofilms, tribocorrosion tests revealed a low friction on titanium covered with biofilms [12, 13]. The properties of the biofilms were similar to those of the lubricant agents used to decrease the wear rate of materials [12, 13]. However, the lowering of pH promoted by microbial species negatively affected the corrosion resistance of titanium surfaces [12, 13]. A wear-corrosion process that takes place during sliding of titanium-based contacting surfaces in a corrosive environment can be a cause of failure in dental implant-supported systems. In dental implant systems, the lower friction in sliding contacts could cause a loss of mechanical integrity of internal connections.

9 Interaction of Wear Debris with Surrounding Tissues

In the case of medical implants and prostheses, wear debris and ions release produced due to the loss of material by bio-tribocorrosion of prosthetic surfaces have been related to tissue inflammatory reactions [108–111, 114]. Additionally, some studies revealed a highly significant relationship between the amount of peri-implant inflammation and the magnitude of alveolar bone loss surrounding implants [110, 111] that can be faster than that surrounding natural tooth due to the absence of inflammatory cellular response provided from periodontal ligament [41].

The presence of metallic ions and particles in human tissues induces the activation of macrophages, neutrophils, and T-lymphocytes with elevation of cytokines and metallic proteinases that can promote bone resorption [115–117]. Coalescence of particles of all classes (including titanium particles) originating from prostheses was often seen in the vesicles of macrophage cytoplasm in the liver (0.1–10 μm in diameter), spleen, and para-aortic lymph nodes [117–120]. In the lymph nodes, titanium particles ranged from 0.1 μm up to 50 μm, while in the liver and spleen the particles ranged from 10 μm [119]. Hallab et al. [121] investigated the binding of metals such as Ti, Co, Cr, Al (originating from implant wear and corrosion) to serum proteins that can mediate immune reactions [120]. Even though the long-term biologic effect of circulating metals is not completely known, it could be determined by the detection and characterization of these metal–protein complexes [120]. After wear tests of titanium alloys in vitro, Okazaki et al. [122] verified a low cellular growth in mediums containing Al and V compared to that in free-Al and free-V mediums. This indicates a potential cytotoxic effect of Al and V for human cells.

A significant release of Ti-, Al-, and V-ions has been reported in previous studies [10]. Literature data have revealed the release of Al- and V-ions caused by passive

film dissolution, though those alloying elements confer good mechanical properties to Ti-alloys. A corrosion of metallic materials has been classified in three classes based on the ion release [109]: (Class I) 10 μg/cm^2 week or less; (Class II) 10–100 μg/cm^2 week or less; (Class III) 100–1000 μg/cm^2 week. Based on Manaranche and Hornberger's [109] study, alloys of class III could stimulate an adverse biological response in patients due the high release of ions. In that respect, CP titanium and Ti6Al4V alloy could induce adverse biological reactions when in contact with high fluoride concentrations [10, 108, 109]. The release of aluminum ions may, however, be considered as a toxic element, while vanadium ions as a mutagenic agent [109].

An association between ultrafine TiO_2 (UF-TiO_2) (<100 nm in diameter) particles and adverse biologic effect has been reported in the literature [94, 108]. Garabrant et al. [94] reported that 50 % of titanium metal production workers exposed to TiO_2 particles suffered from respiratory symptoms, followed by injury of pulmonary function [121]. In agreement with previous studies in rats [123, 120, 124], recent studies in cultured human cells have also shown genotoxicity and cytotoxicity effects of UF-TiO_2 [109]. However, the precise pathways of chromosomal changes, apoptosis formation, and inhibition of cell division by UF-TiO_2 are unclear [109]. These findings led us to consider the possible adverse biological effect of TiO_2 particles (<100 nm in diameter) produced during bio-tribocorrosion mechanisms of titanium in the human body.

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