# Synergistic interactions between corrosion and wear at titanium-based dental implant connections: a scoping review

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## Abstract

Two-piece implant systems are mainly used in oral implantology involving an osseointegrated implant connected to an abutment, which supports the prosthetic structures. It is well documented that the presence of microgaps, biofilms and oral fluids at the implant-abutment connection can cause mechanical and biological complications. The aim of this review paper was to report the degradation at the implant-abutment connection by wear and corrosion processes taking place in the oral cavity. Most of the retrieved studies evaluated the wear and corrosion (tribocorrosion) of titanium-based materials used for implants and abutments in artificial saliva. Electrochemical and wear tests together with microscopic techniques were applied to validate the tribocorrosion behavior of the surfaces. A few studies inspected the wear on the inner surfaces of the implant connection as a result of fatigue or removal of abutments. The studies reported increased microgaps after fatigue tests. Additionally, data suggest that micro-movements occurring at the contacting surfaces can increase the wear of the inner surfaces of the connection. Biofilms and/or glycoproteins act as lubricants, although they can also amplify the corrosion of the surfaces. Consequently, loosening of the implant-abutment connection can take place during mastication. Additionally, wear and corrosion debris such as ions and micro-/nano-particles released into the surrounding tissues can stimulate peri-implant inflammation that can lead to pathologic bone resorption.

***Keywords:*** dental implants, microgaps, bio-tribocorrosion, corrosion, titanium, wear.

## 1. Introduction

The oral cavity represents a challenging environment for any biomaterial. The performance of implant-supported prostheses is dependent on the structural materials used, the design and processing of the materials, as well as on oral environment conditions (1–3). Titanium and its alloys are still the material of choice when it comes to biomedical implanted devices and prostheses. The long-term success rates reported in the literature support the use of titanium-based implant systems due to some properties, such as: high biocompatibility leading to osseointegration, proper mechanical properties on mastication, and corrosion resistance in contact with oral fluids (4–9). However, corrosion of titanium has been detected after exposure to acidic metabolites of the microbial metabolism or even fluoride solutions as found in toothpastes and mouth rinses. Additionally, wear at the implant-abutment contact surface takes place, which in turn can be amplified by the presence of acidic or fluoride-based substances present in the oral cavity (10–14).

Most dental implant systems consist of two main components, named abutment and implant fixture. Those are usually connected by tightening a screw to a specific torque value which is determined by the choice of structural materials and connection design. The implant-abutment connection can reveal a microgap resulting from the mechanical interlocking between the prosthetic abutment and the implant fixture. Microgaps are vulnerable to penetration of oral fluids, glycoproteins and microorganisms. In fact, the implant-abutment connection is filled with these biological materials, which can influence the wear and corrosion of the contacting surfaces (15–17) Additionally, micro-movements of the abutment induced by mastication can lead to friction and wear of the implant-abutment surfaces (18). As a result, over time, the mechanical stability of the implant-abutment connection can be compromised and ions and debris will be released into the surrounding peri-implant tissues (21–24).

The main aim of this focused review was to report on recent *in vitro* and *in vivo* findings on the simultaneous degradation by wear and corrosion of the titanium-based implant-abutment connection in simulated oral conditions or in the oral cavity. This work contributes to a better understanding of corrosion and wear processes at implant-abutment connections and the role of oral fluids and biofilms in those processes. Thus, it becomes a first review on the degradation of implant-abutment connections to the best of our knowledge and therefore that results in insights of great importance in the fields of oral rehabilitation, periodontology, oral implantology and biomedical engineering.

## 2. Dental implant connections

Different dental implant-abutment connections are commercially available for implant-supported prostheses. The osseointegrated implant fixture and the abutment are usually connected by applying a specific torque value to an abutment screw, depending on the design of the structural components. The design of the implant-abutment connection is either hexagon (external or internal) or Morse taper. An excellent fit between abutment and implant establishes an intimate implant-abutment contact leading to a proper mechanical integrity and distribution of masticatory forces. The fit of the implant- abutment assembly is dependent on machining process and properties of the structural materials. However, a poor fit at dental implant-based connections can result in microgaps at implant-abutment connection causing higher displacement of the structural parts on mastication forces or occlusal prematurity from incomplete seating (14,17-20).

The external hexagon connection (Fig. 1A) was the first design developed for dental implant connections (25). Even though an external hexagon connection shows an anti-rotation mechanism inherent to its geometry, the design reveals disadvantages concerning lateral or oblique loading during mastication (19). Many *in vitro* and *in vivo* studies reported the loosening of the abutment screw leading to functional complications (7,18,19,26–28). Thus, cyclic loading from mastication negatively affects the preload of the connection screw resulting in a decrease of the removal torque. This causes an increase in the microgap size and instability of the implant-abutment connection which in turn increases fretting and micromovements of the contacting surfaces in the connection (19,29).

Internal connections (IC) are divided into Morse taper (MT), internal hexagon (IH) (Fig. 1B), octagon or trilobe connections (30,31). MT connections are widely used in oral implantology due to several advantages promoted by an intimate implant-abutment contact (Fig. 1C). Some authors compare MT connections to a cold welding system (32). However, this is not entirely accurate from a micro- or nano-scale point of view. Those connections may or may not have a fastening screw, but in any case the connection and the stability of the conical implant-abutment assembly are dependent on the friction of the contacting surfaces and resultant preload (33–35).

The mechanical integrity of different implant-abutment connections has been studied both *in vitro* and *in vivo* (17-20,29,35–40). Previous studies have shown a significantly higher percentage of mechanical instability in external hexagon implant connections when compared to internal connections (7,38). The long term mechanical integrity of the implant-abutment connection can be evaluated by measuring the removal torque at certain period of clinical inspection (39). The microgaps present at the implant-abutment connection compromise the sealing performance of the MT connection leading to the penetration of substances (e.g. glycoproteins, water, acidic substances) and microorganisms (17,40). Glycoproteins, water and biofilms act as lubricant biological materials decreasing the friction on the titanium surfaces and causing micro-movements of the implant-abutment components (41). Bacterial microleakage can occur in any implant connection, although Morse taper connections generally provide a better fit compared to external hexagon connections (31,39,42,43).

Commercially pure titanium (cp Ti) grade IV is considered the gold standard implant fixture material (5,8) while Ti64Al4V is the most used titanium alloy to produce abutment and prosthetic structures due to its high strength (tensile strength at around 940 MPa). However, there are some concerns regarding degradation and release of ions into the peri-implant area (44). Other titanium alloys has been tested in order to replace Ti6Al4V. On the other hand, the search for more esthetic outcomes has become the main focus of interest in oral implantology. For instance, the grey aspect of titanium-based abutment systems can be noticed in some patients depending on the soft tissue thickness surrounding dental implants. The use of high-strength ceramic abutments (e.g. ytria-stabilized tetragonal zirconia) has been introduced as an alternative structural material for abutments. However, the mismatch in mechanical properties between zirconia and titanium can result in mechanical failure at interfaces (9,30,36,45–49).

## 3. Distribution of stresses at dental implant connections

Stresses, as a result of lateral and vertical forces, are transferred through the tooth root and periodontal ligament to the surrounding bone around natural teeth. The periodontal ligament is mainly composed of collagen fibers which allow a degree of physiologic mobility (50). Additionally, the mechanoreceptors existing in the periodontal ligament can act as an energy-absorbing mechanism on occlusal loading (51,52). When considering implant-supported prostheses, previous studies describe several load/force diagram variations, that depend on several parameters linked to loading, implant design, structural materials and environment (53). Occlusal forces have been reported in a range of 89–150 N in the anterior region, 133–334 N at the canines, 220–445 N at the premolars, and 400–600 N at the molars (1,2,51,52,55). In an implant-supported prosthesis, occlusal forces are distributed through the prosthetic and implant structural materials to the bone tissue. Due to the higher Young modulus (also known as the elastic modulus) of titanium implants when compared to those recorded in bone, absorption of mechanical energy and stress is expected. The Young modulus consists in the relationship between stress (force per unit area) and strain (proportional deformation) that defines the stiffness of a solid material. On occlusal loading, the mismatch in Young Modulus between the titanium-based implant and bone tissue can cause stress shielding at the peri-implant area (1,2,51). Understanding the stress distribution along an implant may help to predict bone resorption as a function of the variation of design, structural materials, implant positioning and bone quality (51). This issue has been studied mainly by finite element analysis estimating the biomechanical behavior of commercial dental implant systems, due to the inherent limitations and time consuming of *in vivo* studies. The highest stress values were reported at the cortical bone and contribute to pathologic bone resorption associated with inflammatory tissue reactions (50,54). Various factors can influence the stress distribution at implant-abutment connections such as loading condition, connection misfit, material properties, connection geometry, contact length and the coefficient of friction (34,39,55). Furthermore, micro-movements occur at the implant-abutment connection during occlusal loading leading to fatigue and wear of the contacting materials (17,40,54). At implant-abutment connections, the abutment screw is subjected to the highest stress concentrations; mainly at the neck region (19,56). Finite element analyses results have revealed a proper gradual stress distribution in the case of internal connections, resulting in less extent of bone volume affected by high loads when compared to those at external connections (38,56–58). In the internal connection, the load is mainly transferred via the internal slope of the implant fixture, as long as the axial compressive force is maintained in the direction of abutment insertion. That also increases the contact pressure and frictional resistance, enhancing the mechanical stability of the implant-abutment connection (20,30,34,35,59). MT connections can maximize the inner contacting implant-abutment area and subsequent friction, thus increasing the preload (31,33,60).

## 4. Peri-implant environment conditions

Similarly to periodontal tissues around teeth, peri-implant tissues comprise gingival sulcus, junction epithelial and connective tissue (32,61,62). The biological width at an osseointegrated dental implant has an average value of 0.16 mm for gingival sulcus, 1.88 mm for junction epithelial, and 1.05 mm for connective tissue (32,63). Such biological width acts like a seal against the penetration of bacteria and their metabolites, as well as corrosive substances and debris from dietary intake (50). The main difference between peri-implant and periodontal tissues is related to the orientation of the collagen fibers and connective tissue attachment to the implant surfaces (43,50). Peri-implant fibers do not have the same type of insertion as those around natural teeth. Collagen fibers run parallel to the implant surface (64), although oblique fibers have been found around MT connections (43).

The apically-directed physiological loss of marginal bone differs from patient to patient and therefore is dependent on implant-abutment features (6,50,65,66). The following causes for pathological bone loss have been proposed: microgaps at implant-abutment connections, remodeling of the biologic width, peri-implantitis, premature load, and implant crest module design (24,32,63,67). It was considered that the degree of marginal bone resorption is directly related to the extent of implant-abutment mismatch (65). Retentive regions like microgaps can accumulate biofilms and therefore stimulate peri-implant inflammatory reactions (32). The biofilm is a complex microbial community organized in an extracellular matrix which gathers substances from food, saliva and microbial metabolism, and is therefore dependent on the nutritional and oral conditions (2). It is suggested that despite the different chemical and physical properties of implant surfaces and teeth, the basic principles of their respective biofilms are quite similar (68,69). Roughness and chemical composition of the surface play a significant role in biofilm formation (2,70,71). Biofilms act as lubricants, decreasing friction between contacting prostetic surfaces (41,70). A mean interfacial discrepancy of about 0.5–60 µm in implant-abutment gaps has been reported in literature (15-18,40,42). Regarding the diameter of microorganisms is less than 10 µm, the implant-abutment gaps can be effortlessly filled by several microorganisms. Also, the penetration of microorganisms in implant internal connections can be caused by microbial leakage at the implant-abutment joints (15,17,40,42). Bacteria have been found on the screw threads and in the apical portion of the abutment screw (15,40,72). The number of peri-implant inflammatory cells increases as the implant-abutment interface depth increases and higher concentrations were found immediately coronal to the implant-abutment connection (24,31). Nevertheless, a lower penetration of microbial cells into Morse taper connections has been reported in previous studies (17,40,42). That can occur due to a higher contacting area between biconical abutment and implant connection surfaces when compared to that in hexagon connections.

Human saliva has an important protective role due to the buffering mechanism due to the presence of a high number of inorganic and organic compounds (73). Nevertheless, several external and intrinsic factors can change saliva pH in the complex environment of the oral cavity (2). The pH value of saliva is normally between 6 and 7 (73) although variations can be noted due to the presence of biofilm, food and dietary or therapeutic substances (74). The buffering mechanism can be limited by a high density of microbial cells or by a low salivary flow rate (1,2,14,70). Then, a lower pH around 3-4 can be found around implant-abutment connections inducing corrosion pathways at the contacting surfaces (1,2,70). On the other hand, the friction recorded on titanium surfaces under sliding against a harder counterbody can be reduced in the presence of water, lipids, and glycoproteins (e.g., mucin) from saliva composition (2,41,82). That can be compared to the effect of commercial lubricant agents (41) and therefore can increase the micromotion of implant-abutment contacting surfaces (17,40) .

## 5. Degradation of titanium-based implant-abutment connections

Titanium and its alloys are considered a gold standard material for biomedical applications due to their physicochemical and biological properties (4,5). The biocompatibility of titanium-based materials is dependent on the titanium surface properties; that is linked to the spontaneous formation of a protective titanium oxide film. This very thin amorphous, low-crystalline and non-stoichiometric titanium oxide film, having a thickness of 1 to 20 nm, is a stable passive layer which protects the surface of the metal from further oxidation (14,75–77). Several previous studies have shown a proper corrosion resistance of titanium-based materials in oral simulation conditions (78,79), although some external substances, such as acidic substances, fluorides, extracellular fluids and lactic acid from the bacterial metabolism, can promote the corrosion of titanium (12,70,76,80–83). However, it should be highlighted that different surface modification methods are currently used to corrode titanium surfaces in the laboratory and therefore to improve the *in vivo* bone tissue compatibility and/or accelerate bone formation. That, in turn, can reduce loading times for dental implants. These approaches are based on the resultant surface layer (77). Porous surfaces composed of thick titanium oxide layers may be produced by anodization of titanium in acidic substances ($H_2SO_4$, $HNO_3$, $H_3PO_4$, and HF). Also, a thick porous Ca/P oxide layer can be produced by *in vitro* anodizing in electrolyte-containing phosphate and calcium-based solutions (2).

A *two-body abrasion* has been reported when two surfaces rubbed away from each other by direct contact with their asperities (1,2). Two-body abrasion can occur in the prosthetic joints and implant-abutment connections surface during masticatory tooth movement (1,2). As a result, plastic flow with metal ejection by plowing and metal detachment forming third bodies (wear particles) takes place. It is important to consider that the inner connection surfaces of the implant and abutment are not completely polished on a microscopic scale (Fig. 2). An attraction between two surfaces that are under relative contact motion takes place after oxide film disruption that results in *adhesive wear*. Wear particles can also be adhered like platelet shapes to surfaces under friction (Fig. 2). However, fractures of the micro-welds resulting from adhesive wear can occur and can increase the wear rate (Fig. 3). During occlusal loading, two-piece implant systems are exposed to vibration and micro-movements causing *fretting wear*. Fretting wear results from repeated loading and unloading cyclic stresses that induce surface breakup, resulting in the loss of material (29,84).

Consequently, the true contact area of the implant-abutment connections is less than the apparent contact, which results in high contact stresses. The transference from one material to the other (Fig. 2), and subsequent loss of material at the connection depends on the hardness, strength and roughness of the structural materials (29,85). *In vitro* studies show a higher wear on the inner surface of the implant than that on the abutment, considering the mismatch in elastic modulus and hardness between the cp Ti implant ($E$ = 105 GPa; 200 HV ) and Ti6Al4V abutment ($E$= 113 GPa; 350 HV) (35,72) (Figure 2). IC titanium abutments with a screw show less wear and plastic deformation than ones without a through bolt screw, showing adhesive wear only in the most coronal portion while the rest of the machined surface is conserved (35). Repeated clinical closing and opening cycles may cause severe wear of the component, gradually decreasing the screw's friction and removal torque values (86).

The metal detachment caused by wear 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 (1,2,12-14,70,81-83). Then, an electrical current flows between anodic (damaged surface) and cathodic (undamaged surface) areas establishing a galvanic cell (1,2,12-14). The oxide layer self-heals immediately after being ruptured in a process called *repassivation* (1,2,12,13,70), with the film on a Ti metal substrate being formed in as little as 30 ms (77) (Figure 3). However, the loss of material increases during the simultaneous wear and corrosion process. In fact, wear, an mechanical phenomenon, and corrosion, an electrochemical phenomenon, have synergistic effects. The combined study of corrosion and wear phenomena occurring in sliding contacts is known as *tribocorrosion*. Than can be defined as a material irreversible transformation induced by the simultaneous action of chemical, mechanical (wear) and electrochemical (corrosion) interactions occurring on surfaces subjected to a relative contact movement (1,2).

Recently, the tribocorrosion behavior of materials is being studied in biological environments (containing proteins living cells) originating the new designation of *bio-tribocorrosion* (1,2,12-14,70,81-83). Mathew et al. (81) reported a negative effect of lipopolysaccharides on the corrosion/wear behavior of titanium in artificial saliva. Additionally, the presence of lipopolysaccharides can induce an accumulation of biofilms. Souza et al (41,70) validated the effect of glycoproteins (e.g. mucin) and biofilms on the bio-tribocorrosion behavior of titanium. Thus, glycoproteins like mucin and biofilms generated an ultra-low friction on titanium under sliding (41). On the other hand, as a result of biofilm growth, a release of acidic substances from carbohydrates metabolism changed the pH and the oxygen content of the local environment (12,70).

## 6. Peri-implantitis induced by corrosion and wear

Peri-implantitis is defined as an inflammatory reaction surrounding dental implants characterized by progressive loss of the crestal bone after the adaptive phase (68,87,88). Meta-analyses estimated weighted mean prevalences for peri-implant mucositis at 43% (CI: 32-54%) and peri-implantitis at 22% (CI: 14-30%) (88). Also, results of the meta-regression showed a positive relationship between prevalence of peri-implantitis and function time and a negative relationship between prevalence of peri-implantitis and threshold for bone loss (88). That further underscore the importance of the review in this manuscript and how important it is to get the word out to both patients and dental care providers regarding these very serious findings.

Only few studies have discussed the presence of titanium wear particles at implant–abutment (24,47,89) connections, as illustrated in Figure 3. Research suggests that metallic ions and particles released from implant-abutment degradation may be involved in cytotoxicity and inflammatory at peri-implant tissues, leading to an increase in sensitivity of gingival epithelial cells to microorganisms (24,92). Those metal ions and debris are recognized as foreign bodies by the immune system, stimulating the migration of polymorphonuclear leukocytes and macrophages, and then activating biochemical mediators associated with bone resorption and peri-implant disease (82,90,91). After wear tests of titanium alloys *in vitro*, Okazaki et al. (97) 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. Thus, the binding of metals such Ti, Co, Cr, Al, V to serum proteins can mediate immune reactions (92,93). 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. Recent studies are bringing new insights on the mechanisms behind peri-implant inflammatory reactions. Ribeiro et al. (98) suggested that $TiO_2$ (anatase form) particles are coated by a bio-complex composed of calcium, phosphorous, hydroxyapatite and organic molecules (proteins). Such bio-complex works as a kind of “Trojan horse” that facilitates nanoparticle internalization by cells. Thereafter, nanoparticles can interact with internal cell organelles and structures, leading to functional modifications and intracellular lesions (98).

An *in vitro* study showed that Ti concentrations of more than 11 ppm have cytotoxic effects, including the induction of necrosis (93). Also, a high release of Ti ions was detected in peri-implant tissues after acidic treatment (e.g. sodium fluoride pH 4.2) which indicates that the fluoride corroded the implant surface under salivary buffering capacity (93). An *in vitro* study evaluated the surface of titanium implants explanted due to peri-implantitis and reported pitting, cracking and fretting-crevice corrosion (94). Fretwurst et al. (23) analyzed bone and mucosa biopsies of patients with peri-implantitis. In 75% of the samples, a higher amount of titanium was detected in some tissue regions. An accumulation of M1 macrophages was found in the region of increased Ti concentration linked to the presence of lymphocytes. Another study reported that 7 of 36 biopsies revealed radiopaque particles (9 to 54 µm in size) of foreign bodies with features of titanium debris mostly surrounding chronic inflammatory infiltrate or a mixture of subacute and chronic inflammation (22). Chronic inflammation can also be seen in foreign body responses in the absence of bacteria (24,95). Progressive bone resorption is related to complex factors resulting in an immuno-osteolytic reaction with the ongoing bone resorption (96). Differences in host response may explain the level of injury progression (87) and therefore it should be highlighted current debates regarding whether to offer implant treatment to patients who suffer from inflammation-related conditions such as diabetes.

Peri-implantitis treatments comprise surgical and non-surgical therapies, of which none are shown to be effective. One of the surgical treatments proposed, named *implantoplasty*, consists of the smoothening of the implant threads that are not surrounded by bone (90,99). Such procedure leaves visible metal debris attached on flap tissues, even after irrigation and acidic decontamination. Even though clinical results show less biofilm retention and a re-accommodation of peri-implant tissues (90,99), the issue requires more research concerning its long-term effects.

## 7.Concluding remarks

Although numerous *in vitro* studies have investigated the mechanical and biological behavior of implant-abutment connections, a few *in vivo* studies have assessed the clinical performance of such connections under dynamic loading and the resulting response of surrounding tissues. *In vitro* studies can not entirely mimic the complex conditions of the oral environment although each critic parameter can be separately evaluated in order to predict long term consequences. In fact, the mechanical instability of implant-abutment connections appears to trigger a chain of synergistic negative events involving issues such as: microgap widening, micromovements of contacting surfaces, wear, ion and debris release, peri-implant inflammatory reactions and peri-implantitis. Such issues are dependent on implant design, structural materials, implant-abutment fit, engineering, manufacturing and individual human conditions. It should be pointed out that a large number of different implant designs and materials have been recently developed. Based on this review, such new implant systems should be thoroughly evaluated both *in vitro* and *in vivo* before being used extensively in clinical practice in order to prevent detrimental effects concerning debris released during the clinical performance of dental implant-supported prostheses. There currently is no effective treatment protocol for peri-implantitis involving surgical and non-surgical therapies. The implant system eventually can fail followed by the potential dangers at the micro- and nano-scale from wear and corrosion. Such consequences of wear and corrosion processes at implant-abutment connections should be further investigated to stimulate novel alternative materials and design for implant-supported prostheses.

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**Figure Legends**

Figure 1. Dental implant connections (A) External hexagon. (B) Internal hexagon. (C) Morse taper.

Fig.2 Schematics of wear at the implant-abutment connection.

Figure 3. Schematics of peri-implant conditions. A. The environment affects the saliva pH. B. Bacterial invasion of the implant-abutment connection