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Research Article

Influence of the biomechanical behavior of TiO_2 and ZrO_2 implants after implantoplasty.

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Abstract: Peri-implantitis occurs around dental implants, and implantoplasty has been used to address this ongoing disease; however, the changes to the physical properties of an implant after implantoplasty have not been well documented. This in vitro study aimed to investigate the effect of implantoplasty on fracture strength and the load required for plastic deformation after fatigue cyclic. Twenty Yttria-stabilized Zirconia implants (ZrO_2 group; Ø 4.1 x 10 mm, Straumann) and commercially pure titanium implants (TiO_2 group; Ø 4.1 x 10 mm, Straumann, control group) implants were placed in polyurethane blocks with 5mm supracrestal. After that, 10 Zirconia and 10 Titanium implants were submitted to implantoplasty in the supracrestal portion using a high-speed handpiece with diamond burs. Then 4 groups (n=10) based on material and implantoplasty treatment were: Ti-Before (titanium implant control group), Zr-Before (zirconia implant control group), Ti-After (titanium implant test group with implantoplasty), and Zr-After (zirconia implant test group with implantoplasty). Each implant received its respective titanium abutment system (Purebase or Variobase), installed with a torque of 35 N.cm using a digital torque wrench. The specimens of each group were submitted to step stress test (30°; 10 Hz; 2 million cycles) until failure and the fracture strength (FS) values were recorded. Two-way ANOVA and parametric comparisons with control were adopted. The FS values for Ti-Bef, Zr-Bef, Ti-Aft, and Zr-Aft were 703.6 ± 68.1 N, 1225 ± 123.8 N, 956.6 ± 86.4 N, and 1035.1 ± 85.7 N, respectively. The number of cycles for Ti-Bef, Zr-Bef, Ti-Aft, and Zr-Aft was 379,323 ± 9,354; 564,779 ± 29,903; 351,451 ± 9,904; 467,009 ± 17,641, respectively. The Ti-Bef and Zr-Bef groups showed a significant FS and a number of cycles than the test groups

($P<0.05$). Both implants investigated (TiO_2 and ZrO_2) showed reduced fracture strength after implantoplasty and cyclic loading.

Keywords: Dental Abutments; Dental Implants; Dental Materials; Titanium; Zirconia.

1. Introduction

The use of dental implants to replace tooth loss is reported as a reliable and well-documented treatment modality. This is one of the most studied and developed treatment options in current dentistry through different forms of presentation, surface treatments, macrogeometry, prosthetic connections, and the materials used for its manufacture.[1]

The biocompatibility of dental implant materials can be discussed as a function of both mechanical and chemical, even the characteristics of the implant materials. [1,2] In this sense, the proper biomechanics of dental implants depends on whether they have the initial strength and resistance necessary to withstand loading within the appropriate functional range. Since mainly the modulus of elasticity of the materials influences the biomechanical behavior of the structure as a whole. [3]

Therefore, the dental implant can be of great importance in situations where the transmission of load to the surrounding bone is key to the long-term survivability of the implant. Despite the recognized biocompatibility of the protective oxide layers on the titanium alloy surface, the release of metal ions has raised concerns in the last decade. [4] As an alternative to alloys commonly used in implant dentistry. [5] Newly developed zirconium (Zr) - and niobium (Nb) - containing titanium alloys and high-strength ceramic alloys such as structural alumina and zirconia have been investigated. [6]

Zirconia has been demonstrated in both in vitro and in vivo experiments to exhibit desirable osseointegration, cellular metabolism, and soft tissue response, primarily resistant to pathogenic microorganisms of periodontal disease (peri-implantitis). Since peri-implantitis has been described as a pathological alteration of the tissues around osseointegrated implants, the microbiota and occlusal trauma are considered its main etiological factors. [7,8] Furthermore, human histology of the zirconia implant demonstrates a suggestive lamina dura morphology and therefore the potential for higher quality osseointegration. [8] From an esthetic point of view, replacing missing teeth with an implant-supported crown in many cases presents a clinical esthetic challenge. In selected cases, especially in the anterior region, the metallic components that appear through the soft tissues can produce an appearance of unnatural graying due to the metallic belt, in this way the zirconia masks. Not least, in conditions of severe recessions, it is able to recover the red and white aesthetics. [9]

All-ceramic implants have been considered as an alternative treatment modality to improve esthetics. Although dental ceramics are biocompatible and esthetic, their brittleness is a concern. [10–13] Furthermore, despite significant improvements in the mechanical properties of structural ceramics, their long-term success is limited by microstructural fatigue, particularly degradation at low temperatures. Recently, zirconium oxide has become one of the most widely used ceramics for restorative dentistry applications. [16–19] However, despite having acceptable osseointegration, the biomechanical strength of zirconium in implant-crown structures has not been tested in long-term clinical trials. term. Unlike orthopedic applications, endosseous ceramic dental implants need to be modified for full crown restoration. This preparation is done using diamond burs. This procedure, as with any abrasive procedure applied to structural ceramics, can result in damage to the surface and associated crack initiation sites. [20,21]

The fundamental aspects of initiation and damage accumulation in endosseous zirconium oxide implants still need to be investigated. The mouth movement stress accelerated life test (SSALT) can provide robust information on the comparative longevity of all-ceramic compared to other systems, especially in conditions where the patient suffers from some periodontal disease, thus often the implantoplasty being an excellent alternative to improve the macrogeometry of the implant. The aim of this study was to test the null hypothesis that there is no influence on the reliability of mouth movement fatigue and failure modes between two-piece ceramic and titanium implants with and without implantoplasty as received and prepared for full crown.

2. Materials and Methods

2.1. Specimen Preparation

Twenty Yttria-stabilized Zirconia implants (ZrO_2 group; Ø 4.1 x 10 mm, Straumann) and commercially pure titanium implants (TiO_2 group; Ø 4.1 x 10 mm, Straumann, control group) implants were placed in polyurethane blocks with 5mm supracrestal. After that, 10 Zirconia and 10 Titanium implants were submitted to implantoplasty in the supracrestal portion using a high-speed handpiece with diamond burs. Then 4 groups (n=10) based on material and implantoplasty treatment were: Ti-Before (titanium implant control group), Zr-Before (zirconia implant control group), Ti-After (titanium implant test group with implantoplasty), and Zr-After (zirconia implant test group with implantoplasty). Each implant received its respective titanium abutment system (Purebase or Variobase), installed with a torque of 35 N.cm using a digital torque wrench) (Figure 1). For the substrate simulation, polyurethane cylinders (F160, Axson Technologies, Saint-Ouen-I`Aumône, France) with a uniform elastic modulus of 3.6 GPa were created inside a 3/4-inch-diameter polyvinylchloride (PVC) tubular section [26]. According to the literature, polyurethane resin is a valid isotropic substrate to simulate bone tissue, in mechanical studies with dental implants, widely applied in in-vitro reports [5,10,24,26,27,28,29,30].

For the substrate preparation, the polyurethane resin was manipulated with equal measures of a base and catalyst until it reached complete homogenization. The resin was poured into the PVC tubes under 45 lbs of pressure in a vacuum pressurizer (Protecni, Araraquara, SP, Brazil) to avoid the incorporation of bubbles. After the resin polymerization, the surfaces were smoothed with SiC sandpaper (#220, #320, #400, and #600) under constant irrigation using an automatic polisher (Ecomet/Automet 250, Buehler, IL, USA) to remove the surface irregularities. A surgical kit (Profile, Titaniumfix, São José dos Campos, SP, Brazil) was used to prepare the perforations in each block. All implants were placed according to the manufacturer's recommendations, with a torque of 35 N.cm, measured with a manual torque wrench (BTG60CN-S model; Tohnichi, Tokyo, Japan). Each implant was placed maintaining 5 mm of the threads exposed above the resin surface, as per ISO for dental implant fatigue (ISO 14801:2016) [31].

To standardize the prosthetic indexing, all specimens were maintained with the anti-rotational lobe toward the buccal face of the crown. In accordance with ISO 14801:2016 [31], aging simulation was performed using hemispherical loading devices, in which the load was applied to a single point of the palatal area in all specimens, avoiding uneven concentrated forces and the premature deformation of specimens

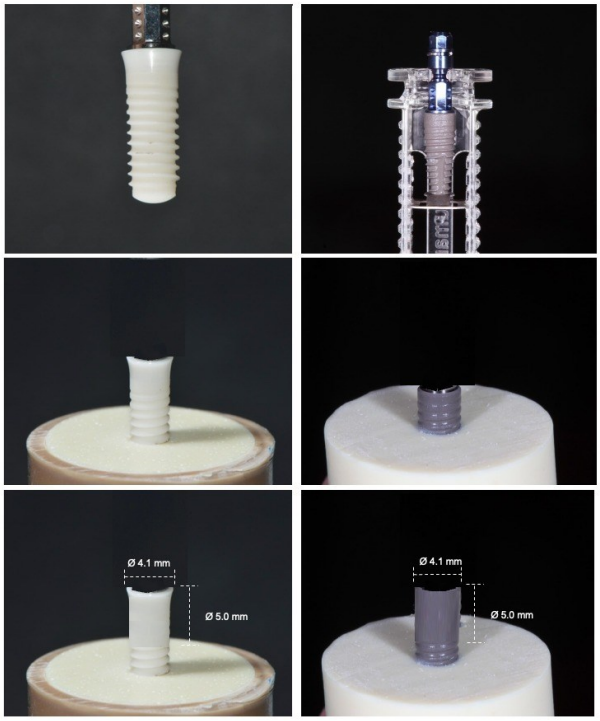

Figure 1. Schematic sequence of the composition of the two experimental groups.

2.2. Maximum Fracture Load

To evaluate the maximum fracture load, half of the specimens were then subjected to a compressive load (0.5 mm/min) in a universal testing machine (Emic DL-1000, Emic, São José dos Pinhais, PR, Brazil) in accordance with ISO 14801:2016 [31].

2.3. Survival Fatigue Analysis

The specimens were then subjected to a 0.5mm/min compressive load in a universal testing machine (Emic DL-1000, Emic, São José dos Pinhais, PR, Brazil) according to ISO 14801:2016 [22]. The specimens were positioned on a base with an angle of 30° about the base of the mechanical fatigue simulator (ER 37,000 Plus, Eros; São Paulo, Brazil) for the fatigue survival test and received 2,000,000 cycles at a frequency of 2 Hz and 200 N load with a 1.6 mm diameter stainless steel applicator, as described in ISO 14801:2016 [22], with the specimens immersed in distilled water at 37 °C. Fatigue resistance analysis was performed using the stepwise test described by (Matos et al., 2022) [23]. The samples were tested in a mechanical fatigue machine (Biocycle, Biopdi, São Carlos, Brazil), with the same device as the monotonic test, inclined at 30°, with a frequency of 10 Hz according to Matos et al. (2022) [23]. Load profiles were analyzed starting at 100 N with the load increasing at each following profile, at intervals of 10000 cycles. The number of cycles and the load at which the specimens fractured during the fatigue test were analyzed by the reliability software SPSS statistics (IBM, Chicago II, USA) using the survival analysis function, Kaplan Meier and Mentel-Cox (Log Rank) ($p<0.05$).

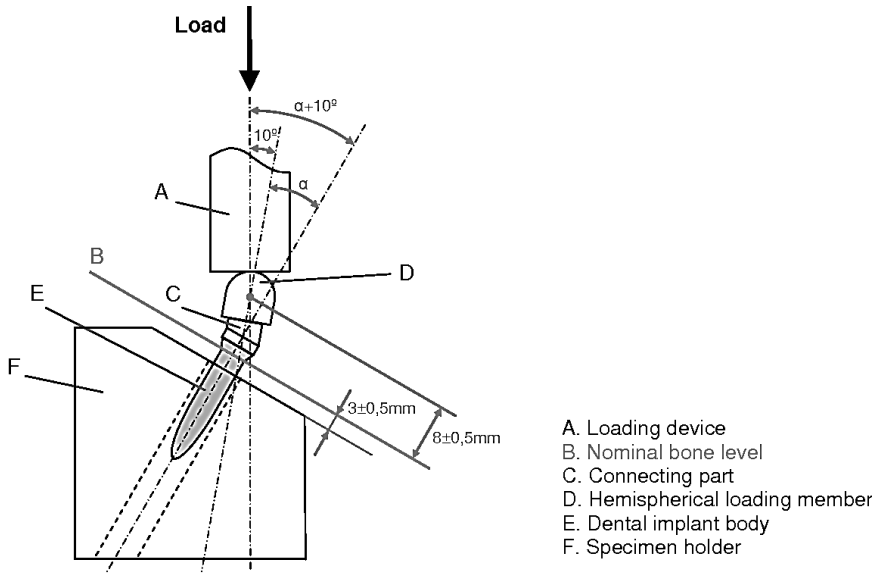

Figure 2. Specimen fixed on a base with 30º of angulation about the ground (ISO 14801:2016), Specimen being submitted to a compressive load of 0.5mm/min in a universal testing machine.

3. Results

3.1. Survival Fatigue Analysis

The stepwise test was also performed due to the absence of failure of any specimen at the end of the 2,000,000 cycles at a frequency of 2 Hz and a load of 200N. Thus, the means and confidence intervals for fracture resistance and cycles to fracture were obtained using the Kaplan-Meier and Mantel-Cox tests (log-rank, 95%), shown in Table 4. Study groups (short and long) were statistically similar in the number of cycles required for fracture (p=0.085), however, in fracture resistance, the long abutment group was statistically superior to the short abutment group (p=0.017).

Table 1. - 1-factor analysis of variance to verify if there is a difference between the maximum load resistance to fracture in both groups.

Group	Mean (Cycles)	SD	CI-Minimum	CI-Maximum
Ti Bef	379323	9354	360988	397658
Ti Aft	351451	9904	332038	370863
Zr Bef	564779	29903	506167	623390
Zr Aft	467009	17641	432432	501587

Group	Mean (Load)	SD	CI-Minimum	CI-Maximum
Ti Bef	856	17.89	821	891
Ti Aft	801	18.98	764	838
Zr Bef	1225	58.93	1109	1340
Zr Aft	1035	35.34	965	1104

The group's survival graphs are presented in Figure 3.

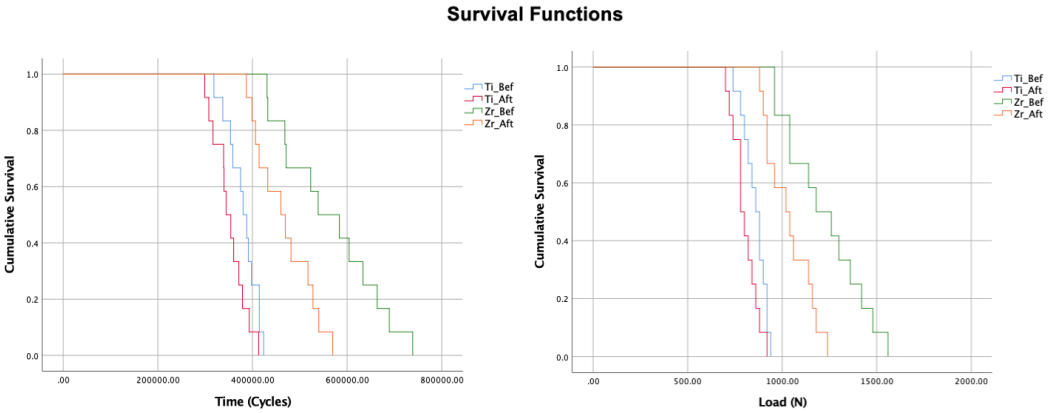

Figure 3. Survival graph of groups as a function of time (cycles) and survival graph of groups as a function of load (N).

Depending on the load or number of cycles, the probability of survival is different among groups. For example, with an 856N, the survival probability of the Ti-Aft is 82%. However, at 1035N the implant has a 76% chance of survival, while the Zr-After group has no chance of survival.

3.2. Statistical Analysis

The specimens of each group were submitted to step stress test (30°; 10 Hz; 2 million cycles) until failure and the fracture strength (FS) values were recorded. Two-way ANOVA and parametric comparisons with control were adopted. The FS values for Ti-Bef, Zr-Bef, Ti-Aft, and Zr-Aft were 703.6 ± 68.1 N, 1225 ± 123.8 N, 956.6 ± 86.4 N, and 1035.1 ± 85.7 N, respectively. The number of cycles for Ti-Bef, Zr-Bef, Ti-Aft, and Zr-Aft was 379,323 ± 9,354; 564,779 ± 29,903; 351,451 ± 9,904; 467,009 ± 17,641, respectively. The Ti-Bef and Zr-Bef groups showed a significant FS and a number of cycles than the test groups ($P<0.05$).

4. Discussion

The results of this investigation determined that FS were significantly reduced after a ZrT implant received implantoplasty and after cyclic loading. Therefore, the null hypothesis was rejected. Most of the studies on the mechanical properties of implants after implantoplasty determine the fracture resistance of the implant body [16,28,29]. It is known that implantoplasty alters the structure of the implant, contributing to its resistance. [30-35]. Wall thickness is reduced and instrumentation can introduce cracks or deformations, facilitating more pronounced deformations, particularly when the material undergoes fatigue in the masticatory environment before the final fracture [16,17,27,29].

Significant mechanical consequences of implantoplasty were only observed after cyclic loading. Test groups after cyclic loading resulted in FS values of (956.6 ± 86.4 N and 1035.1 ± 85.7 N), respectively, compared to control groups (703.6 ± 68.1 N, 1225 ± 123.8 N), respectively. A cyclic loading force of 200 N increasing the load bit by bit is a relatively high amount of cyclic loading when considering the average of the maximum occlusal forces generated. It has been demonstrated that the average occlusal force in the natural dentition varies depending on several external factors, mainly the presence of the number of contacts of the teeth in occlusion or the structure of the implant.

An average maximum occlusal force with dental implant restorations ranges between 50 and 800 N, with an average maximum force of 847 N for men and 597 N for women.43 Although the fracture toughness of the implant used in this study must be beyond the capability of most humans, the force required for fracture after implantoplasty

and cyclic loading is attainable. Implantoplasty may be contraindicated in patients with strong masticatory ability.

The test design used in this study was based on ISO 14801:2016 for dynamic fatigue of dental implants.36 Test parameters based on this standard replicate an extreme clinical scenario where implants with a significant amount of bone loss are loaded 30 degrees off the axle. The ISO test design was followed to maintain the load geometry, but 4mm instead of the specified 3mm of the implant body was exposed to mimic a clinical situation with more extensive peri-implant bone loss. Increased exposure led to an increase in both the moment arm and the bending moment resulting from applied forces, particularly with off-axis loading.

Much of the emerging research, including the present study, suggests that outcomes after implantoplasty are implant-specific and that there is inherent variability in implantoplasty studies and long-term mechanical and clinical outcomes.9,16,21-29 Two important parameters were determined in this study. Cyclic loading simulating material fatigue in a clinical situation was required to determine significant differences in fracture toughness and should be incorporated in future research.

Future studies should examine both final fracture toughness and the s-n curve, as this is a more reliable measure of clinical relevance. Practitioners should consider implant design, implant materials, and patient-specific parameters such as occlusal force when considering implantoplasty as a treatment option. Since, implantoplasty is an excellent treatment in conditions of peri-implant diseases, because it involves the removal of the threads and the polishing of the exposed rough surface of implants that present bone loss. In this sense, more research is needed to evaluate the mechanical effects after implantoplasty, testing various implant designs, materials, and clinical scenarios.

5. Conclusions

Based on the outcomes of this in vitro study, the following conclusions can be drawn:

a) Both implants investigated (TiO_2 and ZrO_2) showed reduced fracture strength after implantoplasty and cyclic loading.
b) The infinite useful life of the evaluated dental implants was well above the threshold of the usual masticatory forces, with the fatigue limit of the implantoplasty group being 801 N (TiO2) and 1035 (ZrO2). Therefore, implantoplasty is demonstrated as a process that increases the survival of dental implants.
c) Implantoplasty does not seem to significantly reduce fatigue resistance, even in unfavorable situations involving normal diameter implants and internal hexagonal connection implants.
d) More studies are needed to determine whether these results are achievable in a real-life clinical setting.

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