Effect of Implantoplasty on Fatigue Performance and Surface Roughness of Narrow Diameter Dental Implants

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Abstract

Implantoplasty (IP) is used in dental implants with peri-implantitis and aims to remove threads and polish rough surfaces in order to prevent bacterial colonization. As a result of this procedure, implant strength might be compromised. We tested 20 tapered screw-shaped Ti6Al4V dental implants with a simulated bone loss of 50%. Ten implants underwent IP and 10 served as controls. Surface topography (S_a , S_z , S_{sk} and S_{dr}) was analyzed with a confocal optical microscope. Subsequently, cyclic loads were applied with a servo-hydraulic mechanical testing machine (5×10^6 cycles at 15 Hz, between the maximal compression force - 529N in the IP group and 735N in the control group - and 10% of that force). We recorded the number of cycles until failure and the type of failure. Implant failure was analyzed by visual inspection and scanning electron microscopy.

Implantoplasty reduced the median S_a from 1.76 (IQR=0.11) to 0.49 (IQR=0.16). The fatigue limits of the control and implantoplasty groups were 551 N and 529 N, respectively. The scanning electron micrographs showed fatigue striations indicating fatigue failure. The infinite life range of the dental implants evaluated was largely above the threshold of usual chewing forces. Implantoplasty seems to render a fairly smooth surface and has a limited impact upon fatigue resistance.

Introduction

Dental implants are a predictable long-term treatment option for the esthetic and functional rehabilitation of patients with partial or total edentulism ^{1,2}. However, different complications can arise and jeopardize the results of implant-prosthetic rehabilitation ³. Peri-implant diseases (both peri-implant mucositis and peri-implantitis) are considered to be the most common long-term complications associated with dental implants. Both disorders are associated to an inflammatory reaction caused by bacterial biofilm, affecting an osseointegrated dental implant ⁴. Peri-implantitis is characterized by inflammatory changes in the peri-implant mucosa and progressive bone loss ⁵.

Nonsurgical treatment has been shown to offer limited efficacy in the remission of peri-implantitis ⁶⁻⁹. A possible reason could be insufficient decontamination of the implant surface, which is exposed to bacterial colonization and is usually moderately rough. In fact, the macro-geometry of the threads and the surface roughness of the implant can further complicate decontamination in the presence of associated peri-implant bone loss ¹⁰. Depending on the morphology and the extent of the defect, as well as on the location of the implant, surgical treatment can involve different approaches and implant surface decontamination techniques ¹¹. Among these techniques, implantoplasty (IP) consists of polishing and smoothing those parts of rough-surfaced implants that are outside the bone contour due to progressive marginal bone loss associated with peri-implantitis, or eventual bone resection during peri-implant surgery ¹². Although this technique has proved effective in clinical studies ^{13,14}, several investigations have reported that IP reduces the fracture resistance of both standard (i.e., 3.75 to 4.5 mm) and narrow (i.e., ≤ 3.5 mm) diameter implants ¹⁵⁻¹⁸. However, none of these studies have determined the fatigue

limit in an unfavorable clinical scenario (i.e., narrow implants with a diameter ≤ 3.5 mm and bone loss equivalent to 50% of their length). These situations are common in dental implants that require IP.

Thus, the primary objective of the present study was to determine the effect of IP upon the fatigue resistance of Ti6Al4V narrow screw-shaped dental implants, with internal connection and a moderately rough surface, in the presence of 50% bone loss. The secondary objective was to analyze the implant surface roughness after IP.

Results

All samples were treated without any deviation from the protocol.

Surface topography analyses

The surface topography results are shown in Table 1. The median S_a (arithmetic mean height), S_z (average maximum height), S_{sk} (skewness of topography height distribution), and S_{dr} (developed interfacial area ratio) values of the IP group were significantly lower than those of the control group ($p \leq 0.001$) (Table 1).

Table 1

Median surface roughness.

	S _a (μm)		S _z (μm)		S _{sk}		S _{dr} (%)	
Sample	Control	IP	Control	IP	Control	IP	Control	IP
1	1.88	0.47	20.86	13.34	0.29	-1.02	18.95	2.62
2	1.84	0.49	19.60	6.20	-0.04	-0.46	18.97	2.57
3	1.82	0.58	30.64	7.66	0.31	-0.86	18.73	2.61
4	1.78	0.44	23.84	7.32	-0.03	-0.78	18.93	2.42
5	1.73	0.65	21.11	15.76	-0.07	-0.70	17.67	5.01
6	1.87	0.50	33.41	10.47	0.64	-0.99	20.96	3.45
7	1.72	0.33	20.15	5.04	-0.05	-1.39	16.29	1.19
8	1.71	0.73	19.21	8.72	0.05	-0.14	16.69	3.84
9	1.74	0.61	19.78	9.09	-0.14	-0.71	16.35	2.95
10	1.72	0.34	27.92	6.31	0.05	0.18	16.89	2.75
Median	1.76	0.49	20.98	8.19	0.01	-0.74	18.20	2.67
IQR	0.11	0.16	8.14	4.16	0.34	0.53	2.26	0.87
P-value	< 0.001*		< 0.001*		0.001*		< 0.001*	
IQR: Interquartile range; IP: Implantoplasty * Significant association (<i>p</i> < 0.05)								

Fatigue testing

Nineteen implants underwent fatigue testing: 10 in the IP group and 9 in the control group (Table 2). Three consecutive samples subjected to IP withstood the 5×10^6 cycles of the initial load level without apparent damage. This was equivalent to 529N, which corresponds to 80% of the maximum compression force ($F_{\max IP}$). The fatigue limit of the control group was 551 N (i.e., 60% of $F_{\max C}$) (Table 2).

Considering that the distance between the nominal bone level and the center of the hemispherical load abutment was 1.3 cm, the maximum bending moments (M) were:

Table 2

Results of the cyclic tests.

% F _{max} total	Peak load (N)	Number of cycles	Failure	
			Location	Description
Implantoplasty group (n = 10)				
95%	628	5,000,000	Absence of failure	
95%	628	5,000,000	Absence of failure	
95%	628	102,360	Implant body	Fracture
90%	595	279,251	Implant body	Fracture
90%	595	5,000,000	Absence of failure	
85%	562	318,799	Implant body	Fracture
85%	562	5,000,000	Absence of failure	
80%	529	5,000,000	Absence of failure	
80%	529	5,000,000	Absence of failure	
80%	529	5,000,000	Absence of failure	
Control group (n = 9)				
80%	735	36,364	Implant body	Fracture
80%	735	66,690	Implant body	Fracture
70%	643	38,830	Implant body	Fracture
70%	643	68,519	Implant body	Fracture
65%	597	112,841	Implant body	Fracture
65%	597	85,644	Implant body	Fracture
60%	551	5,000,000	Absence of failure	
60%	551	5,000,000	Absence of failure	
60%	551	5,000,000	Absence of failure	
F _{max} : Maximum compression force				

Considering that the distance between the nominal bone level and the center of the hemispherical load abutment was 1.3 cm, the maximum bending moments (M) were:

$$(1) M_{IP} = 1.3 \text{ cm} \cdot \sin 30^\circ \cdot 529 \text{ N} = 343.85 \text{ N} \cdot \text{cm}$$

$$(2) M_C = 1.3 \text{ cm} \cdot \sin 30^\circ \cdot 551 \text{ N} = 358.15 \text{ N} \cdot \text{cm}$$

Load versus the number of cycles (S-N curves) in the IP and control groups is represented in Figure 1. Two different regions could be identified: 1) the finite life region was found above 551 N (i.e., 60% of $F_{\max C}$); and 2) the infinite life range which started below that threshold. Similarly, in the load versus the number of cycles plot obtained for the IP samples (Figure 1), we determined: 1) a transition region above 529 N (i.e., 80% of $F_{\max IP}$); and 2) an infinite life range that started below that threshold.

All failed samples exhibited a fracture pattern perpendicular to the longitudinal axis of the implant in a region of the implant body close to the embedding plane (Table 2). This area is the least thickest zone, due to the presence of the hollow space for the prosthetic screw.

Fractographic analysis

The micrographic analysis of the fracture surface revealed a typical brittle intergranular fracture mechanism. In all cases, fatigue failures started at the implant body, with subsequent fracture of the prosthetic screw. More specifically, the fracture began on the side of the implant subjected to continuous and oscillating stresses. Accumulated damage led to rupture on exceeding the mechanical resistance of the material (Figure 2).

Discussion

The present *in vitro* study assessed the reduction of fatigue strength of narrow-diameter dental implants with internal hexagonal connection in a model that simulated a horizontal peri-implant defect equivalent to 50% of the implant length. To the best of our knowledge, this is the first study to analyze the effect of IP in this worst-case scenario. Although implantoplasty significantly reduced the fatigue limit, the failure threshold was still above the usual chewing forces^{19,20}. In addition, changes in surface roughness were evaluated by confocal microscopy. Implantoplasty significantly reduced the roughness of the implant surface to a minimally rough, groove-free surface – the resulting roughness being comparable to that of a machined implant surface.

In our study, IP was carried out under conditions that simulated the real-life clinical scenario, though less challenging. As previously reported with a similar protocol²¹, roughness zones might be difficult to reach in clinical practice, particularly in areas with difficult access or when it is not possible to remove the prosthesis. This might result in more aggressive thinning of the implant walls and, consequently, poorer mechanical properties than those reported herein.

One of the major concerns related to IP is the mechanical behavior of the dental implant after polishing²². Although Schwarz et al.²³ recommended that IP should be limited to the threads of the implant, a reduction of the thickness of the implant walls is to be expected^{18,21,24–28}. Aside from the dimensions of

the implant walls, other factors influence the mechanical behavior of dental implants: the implant material ²⁹, the implant-abutment connection design ^{30,31}, implant diameter ^{15,18,32,33}, crown to implant ratio ²⁵, crown height ³⁴, and nominal bone level ²⁶. This is the reason why we selected a worst-case scenario involving commercially pure titanium narrow diameter (< 3.5 mm) implants, with an internal connection and thin walls, an unfavorable crown-to-implant ratio, and with a significant loss of supporting bone.

In a previous study ²¹ we performed compression tests and found IP to significantly reduce fracture resistance ($p < 0.001$). Specifically, the external hexagonal, internal hexagonal and internal conical connection groups exhibited a decrease in $F_{\max}$ of 27.96%, 28.00% and 29.41%, respectively ²¹. However, the clinical relevance of these static loading tests is limited, because factors such as time or environment are not taken into account ³⁵. In fact, mechanical failure of dental materials usually occurs once they withstand ³⁶ repeated cycles of low-energy stress, rather than higher static loads. Three *in vitro* studies have analyzed the effect of cyclic loading upon the fracture resistance of implant materials ^{18,27,37}. However, to the best of our knowledge, the present study is the first to determine the effect of IP upon the dental implant fatigue limit. The application of $> 1 \times 10^6$ cyclic loads reduces fracture strength by introducing a "mechanical aging" effect in the tested components ^{27,37}. Our results suggest that even in this worst-case scenario (i.e., 3.5 mm diameter internal hexagonal connection implant, with bone loss equivalent to 50% of its length and subjected to IP), implants showed an infinite life range above 500 N, which is well above the threshold of the forces recorded during chewing and swallowing (around 250 N) ^{19,20}.

Several IP protocols have been described in the literature. Most publications agree that IP, regardless of the burs used, significantly reduces roughness ³⁸. This could have an impact upon peri-implant health, as the composition and development of biofilms on the surface of dental materials correlate with their surface roughness and free surface energy ³⁹. Other physicochemical properties, such as surface charge or substrate stiffness, appear to be of lesser importance ^{40,41}. Regarding the surface roughness of the dental implant, R_a and S_a are the most appropriate parameters for predicting susceptibility to bacterial adhesion ⁴². In fact, roughness has no influence upon bacterial adhesion at $R_a < 0.2 \mu\text{m}$ ^{43,44}. Although our IP protocol resulted in a minimally rough surface ($S_a = 0.49 \mu\text{m}$), this still might not be smooth enough to impede bacterial adhesion. Thus, even if complete decontamination of the surface is achieved with IP, bacterial recolonization will occur within a short period of time. Therefore, it is crucial for the prosthesis to facilitate hygiene of the treated area and, at the same time, for the patient to maintain good plaque control. Furthermore, the patient should follow a maintenance program with follow-up visits at least every 4-6 months in order to avoid reinfection or the recurrence of peri-implantitis ⁴⁵.

Conclusions

The infinite life range of the evaluated dental implants was largely above the threshold of usual chewing forces, with the fatigue limit of the implantoplasty group being 529 N. Thus, implantoplasty does not seem to significantly reduce fatigue resistance even in unfavorable situations where narrow-diameter internal hexagonal connection implants are involved. With carbide burs and silicon carbide polishers, S_a values $< 0.5 \mu\text{m}$ can be obtained. Further studies are required to determine whether these results are achievable in the real-life clinical setting.

Methods

Twenty tapered, screw shaped Ti6Al4V (titanium grade 5) commercial dental implants were tested in the present *in vitro* study (Biomimetic Ocean® 3.5 mm wide and 10 mm long with internal hexagonal connection, Avinent® Implant System, Santpedor, Spain). The surface was moderately rough after a sandblasting, acid-etching and anodization process. We used a computer-generated random sequence to allocate 10 implants per group, and subsequently performed implantoplasty of the implants in the IP group.

Implantoplasty procedure

Implantoplasty was performed following the simplified three-bur protocol described by Costa-Berenguer et al.²⁴. We inserted a cover screw to protect the implant connection from titanium debris, and removed the threads of the coronal half of the implants using an oval-shaped tungsten carbide bur (H379.314.023 Komet, GmbH & Co. KG, Lemgo, Germany) with an air-driven high-speed handpiece under water irrigation. Then, we polished the resulting surface with two silicon carbide polishers and the same handpiece (9618.314.030 and 9608.314.030 Komet, GmbH & Co. KG, Lemgo, Germany) (Figure 3).

Implantoplasty was performed by an experienced clinician (O.C-F.) under loupe magnification until the 5-mm coronal portion of the implant exhibited a uniform smooth and shiny surface. The pressure applied and the number of strokes were not standardized in order to increase the external validity of the study. A new set of burs or tips was used for every other implant. After IP, the implants were thoroughly cleaned by irrigation using distilled water and dried with compressed air. Finally, the cover screw was removed.

Surface topography analyses

All samples were analyzed with a confocal optical microscope (Leica® DCM 3D, Leica Microsystems AG, Wetzlar, Germany) under 20x magnification. We determined three regions of interest of $600 \times 450 \mu\text{m}$: immediately below the smooth surface of the platform (T_0), at 2.5 mm ($T_{2.5}$), and at 5 mm (T_5) from the platform in the apical direction (Figure 4). The images were processed using the LeicaMap® application (Leica Microsystems AG, Wetzlar, Germany).

The surface texture of each area was defined using the following normalized three-dimensional parameters: S_a (arithmetic mean height), S_z (average maximum height), S_{sk} (skewness of topography height distribution) and S_{dr} (developed interfacial area ratio). Form was previously removed, and a

Gaussian filter of 30 μm was applied for roughness and waviness. Only roughness parameters were assessed.

Cast preparation

In a second step, we embedded the implants in the same position using resin casts, in such a way that 5 mm of rough surface was exposed. This approach was chosen to simulate a horizontal bone resorption of 5 mm (50% of the total implant length), which is 2 mm more than the International Standardization Organization (ISO) 14801:2016 specifications. The epoxy resin was EA 3471 A and B Loctite® (Henkel AG & Company, Düsseldorf, Germany) to simulate bone (Young's modulus of elasticity $\geq 3 \text{ GPa}$).

Fatigue testing

We carried out fatigue testing in room air and at room temperature using a servo-hydraulic mechanical testing machine (MTS Bionix 370, MTS®, Eden Prairie, USA) equipped with a 15 kN load cell (MTS Load Cell 661.19H-03, MTS®, Eden Prairie, USA). We screwed identical hemispherical abutments to each implant with the torque recommended by the manufacturer (35N·cm). The loading center was located 13 mm above the resin (nominal bone level). We placed the samples in a stainless-steel clamping jaw so that loading had an angle of 30° to the longitudinal axis of the implant (Figure 5).

Each specimen received a maximum of 5,000,000 cycles of a uniaxial load, perpendicular to the tangent of the dome of the hemispherical abutment. Loading range was between a maximal nominal value and 10% of this value ($R = 0.1$). To minimize the vibrations of the testing machine, sinusoidal load frequency was kept at 15 Hz. We used TestStar II® software (MTS®, Eden Prairie, USA) to record data in real time.

In accordance with ISO 14801:2016, tests were carried out applying a minimum of four series of loads, the first of which was equivalent to 80% of the maximum compression force (F_{maxC} and F_{maxIP}), which was determined in a previous study to be 735 N and 529 N for the control and IP samples, respectively ²¹. At each load level, two samples were evaluated, considering 5×10^6 cycles as an infinite life criterion. If any of the samples collapsed before reaching the specified number of cycles, the procedure was started again with two new implants and under a lesser load (20% if $\geq 60\% F_{\text{max}}$ and 10% if $< 60\% F_{\text{max}}$). When two consecutive samples reached 5×10^6 cycles without failure, an additional test was performed with a third sample. If the latter did not fail (i.e., 3 consecutive samples without apparent failure), this point was considered to be the fatigue limit beyond which the implant could withstand an infinite number of loading cycles. In case the fatigue limit was reached in less than four load series, additional levels (1, 2 or 3) were established by applying a load 5% higher than the previous one. The number of cycles and the state (i.e., intact or failed) of each tested specimen were recorded. Failure was defined as the elastic limit of the material, permanent deformation, loosening of the implant assembly, or fracture of any component.

Additionally, for the maximal supported load, the maximal bending moment (M) was calculated using the equation (3):

$$(3) M = l \cdot \sin 30^\circ \cdot F$$

Where l is the distance (in cm) from the center of the load hemisphere to the nominal bone level and F (in N) the maximal supported load.

The results of the fatigue tests were displayed in a load versus number of cycles plot (i.e., S-N curve or Wöhler's curve), which represents the number of load cycles of each sample (logarithmic scale) and the corresponding maximal load (linear scale).

Fractographic analysis

All failed specimens were assessed by visual inspection and scanning electron microscopy (SEM; Quanta-200, Field Electron and Ion Company, Hillsboro, USA) to describe the failure pattern.

Statistical analysis

Categorical variables were reported as absolute and relative frequencies. We explored normal distribution of scale variables (roughness parameters) with the Shapiro-Wilk's test and visual analysis of the P-P and box plots. The mean and standard deviation (SD) were calculated and, if a normal data distribution was ruled out, the median and interquartile range (IQR) were calculated. The Mann-Whitney U-test was used to compare the groups. The statistical analysis was carried out with the Stata14 statistical package (StataCorp®, College Station, TX, USA) at a level of significance $p < 0.05$.

Declarations

DATA AVAILABILITY: Datasets generated and analyzed in the current study are available from the corresponding author on reasonable request.

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AUTHOR CONTRIBUTIONS

O.C-F, E.V-C. R.F., C.G-E. conceived and designed the study; C.G-E., E.V-C. supervised the study; O.C-F, A.J-C. acquired and analyzed the data; O.C-F, J.T-S. performed the statistical analysis; O.C-F J.T-S. R.F., E.V-C. interpreted the data and were involved in preparing the manuscript. All authors reviewed the final manuscript, approved the submitted version and agreed both to be personally accountable for the author's own contributions and to ensure that questions related to the accuracy or integrity of any part of the work are appropriately investigated, resolved, and the resolution documented in the literature.

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Figures

Figure 1

Load versus the number of cycles plot (S-N curve) for each group.

IP: Implantoplasty; NF: Number of load cycles (i.e., 5×10^6); LF: Fatigue limit.

Figure 2

Fractographic analysis with scanning electron microscopy. a) Control sample at 70x and 1000x; b) Test sample at 70x and 1000x. Red arrows indicate the presence of fatigue striations.

Figure 3

Simplified three-bur protocol IP procedure. a) Macroscopic appearance of the implant; b and c) Macroscopic appearance of the implant after applying the tungsten carbide bur; d-e) Macroscopic appearance of the implant after applying the silicon carbide polishers.

Figure 4

Confocal microscopy surface topography analyses. a) Control group sample; b) IP sample. CV: Area of interest.

Figure 5

Schematic representation of the test setup according to ISO 14801:2016, except for bone nominal level.