The Implant-Abutment Interface of Alumina and Zirconia Abutments

Bulem Yüzügüllü, DDS, PhD;* Mehmet Avci, DDS, PhD†

ABSTRACT

Background: Although ceramic and titanium abutments are widely used in clinical practice, the mechanical characterization of the implant-abutment interface for ceramic abutments has not been evaluated after the dynamic loading.

Purpose: The purpose of this study was to assess the implant-abutment interface after the dynamic loading of titanium, alumina, and zirconia abutments.

Materials and Methods: Fifteen aluminum oxide, zirconium oxide, and titanium abutments were manufactured by the Procera System® (Nobel Biocare AB, Göteborg, Sweden) and were connected to Ø 3.75 × 13-mm regular platform implants (MK III, Nobel Biocare AB) secured in a 30° inclined plane. A mechanical testing machine applied compressive dynamic loading between 20 and 200 N at 1 Hz on a standard contact area of copings cemented on abutments for 47.250 cycles. The measurements of microgaps at the implant-abutment interface from the labial, palatinal, mesial, and distal surfaces of each specimen were undertaken by scanning electron microscope analyses prior to and after the experiments. The data of the microgaps before and after the dynamic loading were statistically assessed using the Wilcoxon signed rank test and the Kruskal–Wallis variance analysis ($\alpha = 0.05$).

Results: Coping fracture, abutment fracture, or abutment screw loosening or fracture was not detected in any specimen during the entire test period. After the dynamic loading, the titanium abutment control group revealed an increased microgap (3.47 µm) than zirconia (1.45 µm) and alumina (1.82 µm) groups at the palatinal site ($p < .05$). The mean measurement values at different measurement sites of specimens within and between each abutment group were similar ($p > .05$).

Conclusion: Owing to their comparable microgap values at the implant-abutment interface after the dynamic loading, ceramic abutments can withstand functional forces like conventional titanium abutments.

KEY WORDS: ceramic abutment, implant-abutment interface, dynamic loading

INTRODUCTION

The long-term success of osseointegration is largely determined by the maintenance of implant–bone interface.[1] Until the recent introduction of ceramic abutment systems, abutments were generally made of machined titanium designed to have various machining tolerances at the interface between the implant and the abutment.[2] The use of standard cylindrical titanium abutments occasionally posed aesthetic difficulties for maxillary anterior single-tooth implant restorations and did not allow the correction of implants placed in nonaxial orientation to the desired placement of the restoration.[1] The use of ceramic abutments as a support for the missing single tooth with a dental implant in the anterior region of the mouth has been successfully used clinically since the early 1990s.[3] The introduction of aluminum oxide or zirconium oxide implant abutments, allowing individualization by milling, provided new opportunities for single-tooth implant restorations in the aesthetic zone. These abutments are distinguished by their tooth-matched color, good tissue compatibility, nontoxicity, and intrasulcular adaptability.[3–6]

*Research fellow, Department of Prosthodontics, Faculty of Dentistry, Hacettepe University, 06100 Sıhhıye, Ankara, Turkey; †professor, Department of Prosthodontics, Faculty of Dentistry, Hacettepe University and private practice, Istanbul, Turkey

Reprint requests: Dr. Bulem Yüzügüllü, DDS, PhD, Department of Prosthodontics, Faculty of Dentisty, Baskent University, 11. Sok. No:26 Bahcelievler, 06490 Ankara, Turkey; e-mail: bulemy@gmail.com

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The specific problem for titanium transmucosal implant-abutment systems, particularly in butt–joint or external–internal hex designs, is that the microgap between the implant and the abutment may increase because of bending moments and consecutive fatigue and wear at the interface. This is followed by plaque retention at the interface, resulting in clinical sequelae such as bone loss, peri-implantitis, and possible loss of osseointegration.[7–10] Jansen and colleagues[8] used 13 different two-piece implant systems to investigate the correlation between component fit and microbial leakage of the implant-abutment interface. The authors concluded that the implant-abutment systems with fine mating surfaces may not prevent microbial leakage. Piattelli and colleagues[10] compared cement-retained and screw-retained implant-abutment connections by scanning electron microscope (SEM) analysis and concluded that cement-retained abutment implants offer better outcome in terms of fluid and bacterial permeability. Further, Dellow and colleagues[11] reported that manufacturing variations can result in as much as 0.1 mm of space at the implant-abutment interface. Therefore, the authors evaluated implant-abutment interface fit using SEM analysis of four implant systems and interchanged components between the various systems. According to a study by Callan and colleagues,[12] a retrospective review and evaluation of 350 dental implants in 203 patients and SEM examination of 45 failed implants indicates that the implant-abutment interface, when in subgingival position, is likely to harbor bacteria. These results were associated with circumferential crestal bone loss of more than 3 mm. Hermann and colleagues[13] stated movement between the implant and the abutment may be implicated in contributing to crestal bone loss. In addition, there are several studies investigating the microgap between different types of titanium abutments and implant systems produced by different manufacturers.[7,8,10–22] However, changes occuring at the implant-abutment interface of titanium, alumina, and zirconia abutments subjected to dynamic loading has not been evaluated so far. The objective of this in vitro study was, therefore, to assess the changes in the implant-abutment interface comparing titanium, aluminum oxide, and zirconium oxide abutments placed on Brånemark® (Nobel Biocare AB, Göteborg, Sweden) implants subjected to a standard dynamic loading regimen and evaluated by scanning electron microscopy analysis.

MATERIALS AND METHODS

Implants and Fabrication of Abutments and Copings

Fifteen Ø 3.75 × 13-mm regular platform implants (MK III, Nobel Biocare AB) were used in the present study. Two test groups (aluminum oxide and zirconium oxide) and one control group (titanium) abutments, all consisting of five specimens were fabricated by the Procera System (Procera, Nobel Biocare AB).

During the preparation of the copings, a preparation of a maxillary central incisor was modeled with hard wax simulating a 1.5-mm axial and 2-mm incisal reduction with 1-mm width chamfer-type 360° margin and 6 to 10° axial wall taper according to an average-sized incisor preparation.[23] To standardize the loading area throughout the investigation, a loading area of 3 mm^2 was prepared in the middle third of the palatinal surface of the zirconia copings above the cingulum area of the preparation. The reason for this was that the vertical loading stylus in the loading apparatus had a 3-mm^2-diameter tip. The prepared pattern was scanned (Procera Scanner, Nobel Biocare AB); zirconium oxide copings were fabricated to fit the respective abutments (Procera facility, Sandvik, Stockholm, Sweden).

Then, the abutments were secured to the implants by applying 32 N·cm torque to titanium abutment screws with a manual torque wrench (Brånemark). Before the cementation of the copings, the aluminum oxide abutments were abraded with 50-μm aluminum oxide. All abutments and copings were steam-cleaned and dried for surface preparation. The copings were cemented using a dual-polymerizing adhesive resin luting agent (Bifix DC, Voco, Cuxhaven, Germany) according to the manufacturer's instructions. Manual pressure was initially applied to set each coping on the abutments. Then, each specimen was placed in a custom-made vertical static loading apparatus for 5 minutes under 2-kg static load and bench set for 24 hours.

Scanning Electron Microscopy

To measure the microgap between the implants and abutments, before and after the loading experiments, a SEM (JEOL JSM-5600 LV, JEOL, Tokyo, Japan), digital image software (Voyager EDS R 3050, Noran Instruments, Inc., Middleton, WI, USA) and analyzer (Voyager EDS R 3050, Noran Instruments, Inc.) were used. As

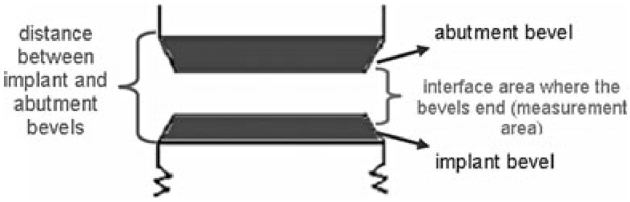

Figure 1 The diagram of the measurement area between the implant-abutment interface.

gold would lead to an additional thickness in the marginal gap area, the specimens were not gold sputtered before SEM observations. Instead, the specimens were cleaned with ethyl alcohol in an ultrasonic cleaning bath (Amsco, Reliance Sonic 250, Steris Corp., Mentor, OH, USA) for 5 minutes and placed in dry sterilization packs. This cleaning procedure was repeated prior to the SEM evaluation of the specimens after the loading experiments were completed. The marginal gap of each specimen before and after the dynamic loading was measured at eight locations with a magnification of ×4,000 in a viewing angle perpendicular to the long axis of each implant-abutment complex. The implants were turned in 90° increments starting from the labial surface, and two points of measurement were randomly selected on each surface. The rounded edge of the Brånemark implants and abutments was not included in the measurements (Figure 1).

Setup and Dynamic Loading

A custom-made jig, consisting of two parts, was fabricated for placing the implants on the load frames of the loading machine. The first part consisted of a vertical loading stylus. The second part was a split (two-part) 30° inclined plane to simulate the functional stresses along the central incisor root angulation, which also had a socket to secure the implant (Figure 2A).[24] The specimens were secured into this socket by autopolymerizing methyl methacrylate acrylic resin (Palavit G Cold-Curing Resin, Heraeus, Kulzer GmbH & Co., Hanau, Germany), which exhibits an elastic modulus similar to that reported for trabecular bone (1.95 GPa).[25] On the 30° inclined plane, the acrylic resin embedded implants were fixed in place by four screws, allowing a sliding movement within slots for each of the 15 specimens (see Figure 2B). A universal testing machine (Instron 8516 Plus Universal, Instron Corp., High Wycombe, UK) was used to apply a cyclic load to the specimens. A dynamic loading between 20 and 200 N[15] was applied to the 3-mm^2 loading area of each abutment-coping complex for 47.250 (45 days) cycles. Force application was not randomized and was cyclically ramped between two limits. A sinosoidal waveform at 1 Hz was applied to simulate values found in human mastication.[26] The specimens were tested in a dry environment.

Statistical Analysis

Within- and between-group comparisons for microgap values before and after the dynamic loading were evaluated by Kruskal–Wallis variance analysis, and the Wilcoxon signed rank test was used to statistically assess the effects of the experimental factors ($\alpha = 0.05$).

RESULTS

The microgaps of abutments before and after the dynamic loading are presented in Table 1. The values of the microgap measurements from the labial, mesial, distal, and palatinal surfaces of each specimen are

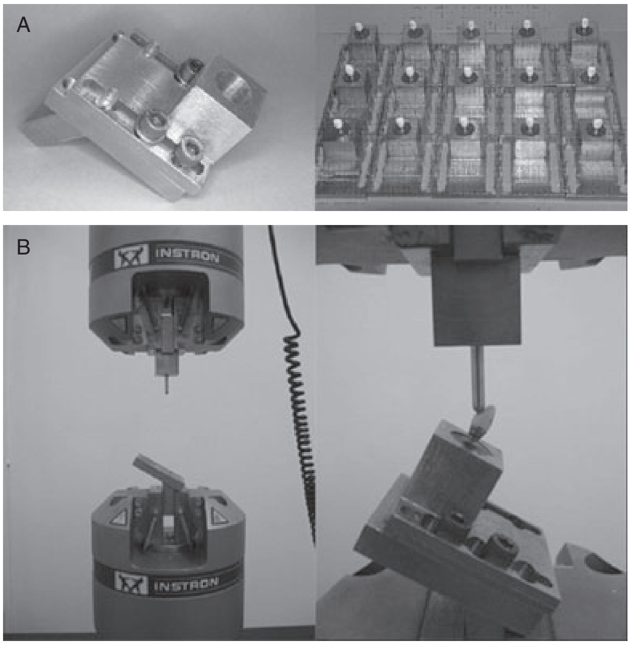

Figure 2 *A*, The split 30° inclined plane. The upper part of the jig is assembled to the lower part by four screws (left). The inclined section of the lower part is attached to the universal testing machine, which allows 30° oblique load application. Note also the 15 implants with abutments and copings in place secured to the upper component of the jig (right). The slots at the two sides of the upper part are used to slide the upper part on the lower part to provide a precise fit of the loading probe on the loading area of the coping, and also, ease in assembling the jig by four screws. *B*, The universal testing machine without implants and the loading probe in place (left). Application of dynamic loading to the implant-abutment-coping system (right).

TABLE 1 Mean Microgap Measurements (μm) of Abutment Groups before and after Dynamic Loading

	Al		Zi		Ti	
Specimen #	BL	AL	BL	AL	BL	AL
1	3.665	1.588	2.883	1.423	2.339	2.371
2	3.621	2.120	2.355	1.380	4.012	4.971
3	2.363	2.635	2.212	1.649	3.091	2.319
4	2.478	3.191	1.994	2.607	3.328	2.767
5	3.650	2.930	3.157	2.239	3.209	2.686
Mean (SD)	3.15 (0.67)	2.49 (0.64)	2.52 (0.48)	1.85 (0.54)	3.19 (0.59)	3.02 (1.10)

AL = after loading; Al = aluminum oxide; BL = before loading; Ti = titanium; Zi = zirconium oxide.

presented in Table 2. The SEM images of three abutment groups, in ×200 and ×4,000 magnification, are presented through Figures 3–5 and Figures 6–8, respectively.

During the entire test period, coping, abutment, or abutment screw loosening or fracture was not detected in any specimen. Within- and between-group comparisons did not reveal significant differences for unloaded specimens ($p > .05$). After the dynamic loading, microgaps were similar at labial, mesial, and distal surfaces in all groups ($p > .05$), but higher for titanium abutments than other groups at the palatinal site ($p < .05$). Within-group comparisons of loaded specimens did not reveal any differences, except for the titanium abutment group, which showed higher values on the palatinal site compared to the labial site ($p > .05$). The gap of zirconium oxide abutments was relatively lower than those of aluminum oxide and titanium abutment groups before and after loading, but the difference between groups was statistically insignificant ($p > .05$).

DISCUSSION

Studies on the comparison of microgaps between implant-abutment interfaces demonstrate different results because of the differences in machining tolerances of implant systems as well as measurements using SEM analysis.[8,11] The differences due to measurements using SEM analysis depend mostly on whether or not the distance between the implant and abutment bevels is included in the calculation of the gap space measurements. Including the distances between bevels results in larger microgap values leading up to 36 to 86 μm as reported in the study by Byrne and colleagues.[1] Excluding the bevels in the measurement, which would not influence the interface between the implant and abutment,[8,11] results in smaller microgap values as was seen in the present study.

During the transmission of masticatory forces via the restoration-abutment interface to the dental implants, the lateral component of force is thought to be responsible for creating bending moments.[20,21] On the surface facing the external load, the implant and the abutment experience tensile stress from bending, while on the opposite surface, the connection is subject to compression.[14] According to Kohal and colleagues,[17] the nonaxial forces effecting the anterior maxilla cause higher stress concentrated along the facial and lingual surfaces of the implant-abutment interface. In the present study, it was for this reason that the dynamic

Table 2 Mean (μm) and SDs of Two Points of Microgap Measurements from Labial, Mesial, Distal, and Palatinal Surfaces of Three Abutment Groups before and after Dynamic Loading

	Labial		Palatinal		Mesial		Distal	
	BL	AL	BL	AL	BL	AL	BL	AL
Al	2.79 (1.46)	2.91 (0.99)	3.47 (1.69)	1.82 (0.26)	3.22 (1.02)	1.89 (0.38)	1.65 (0.52)	1.85 (0.44)
Zi	1.87 (0.39)	1.55 (0.65)	2.70 (1.01)	1.45 (0.48)	2.01 (0.86)	1.51 (1.13)	1.99 (0.68)	1.42 (0.25)
Ti	2.53 (0.83)	2.15 (0.89)	3.42 (2.27)	4.46 (2.83)	2.95 (0.92)	1.91 (0.76)	2.38 (0.41)	2.08 (0.76)

AL = after loading; Al = aluminum oxide; BL = before loading; Ti = titanium; Zi = zirconium oxide.

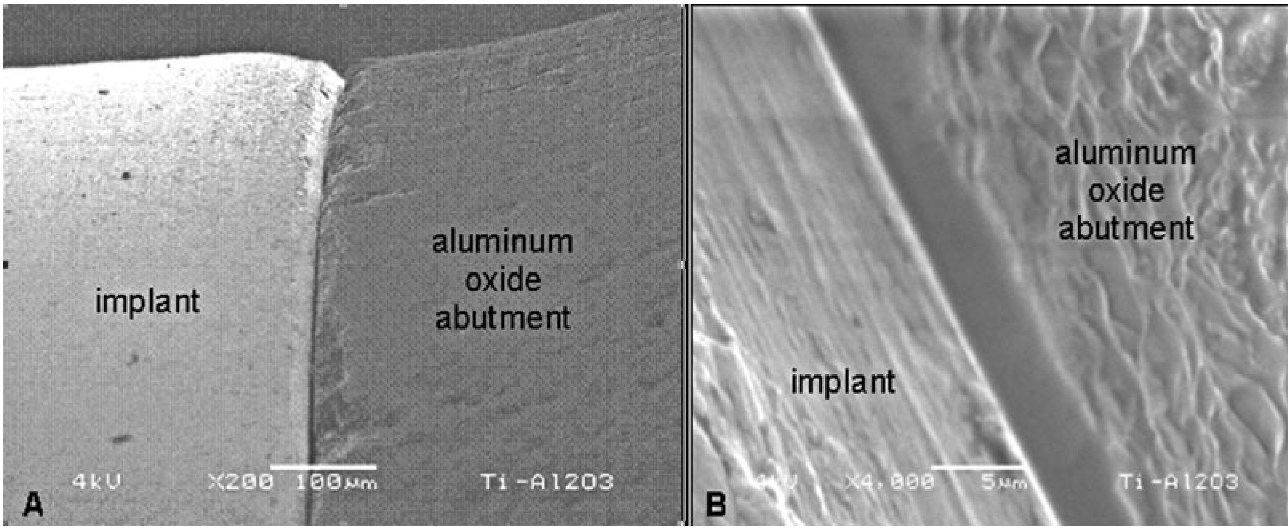

Figure 3 *A* and *B*, Scanning electron microscope photographs of aluminum oxide abutment group before the dynamic loading.

loading was performed as 30° oblique loading, which is more relevant clinically, and could therefore better simulate the mechanical events occurring at the implant-abutment interface of butt–joint designs.

Another important factor significantly influencing the mechanical characterization of an implant system is the implant-abutment mating design. In the external hexagonal configuration used in the present study, the axial preload of the abutment screw is the most important factor in maintaining stability at the implant-abutment interface. This screw alone secures the abutment during horizontal and oblique loading. However, there is no form of lock or positive locking by the external hexagon, which determines the rotational position but does not absorb any lateral loading. The optimal preload corresponds theoretically to the yield point of the screw. The purpose of torque tightening of the titanium abutment screw in general clinical practice as well as in the present study is to achieve the optimum preload that maximizes the fatigue life, while offering a

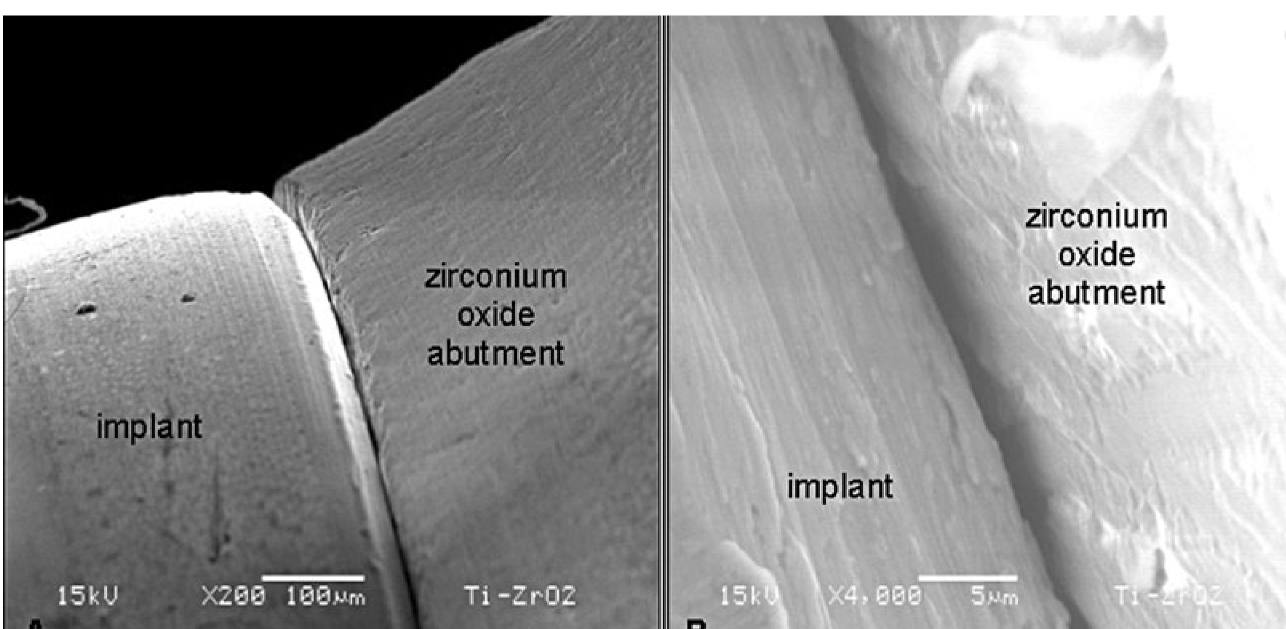

Figure 4 *A* and *B*, Scanning electron microscope photographs of zirconium oxide abutment group before the dynamic loading.

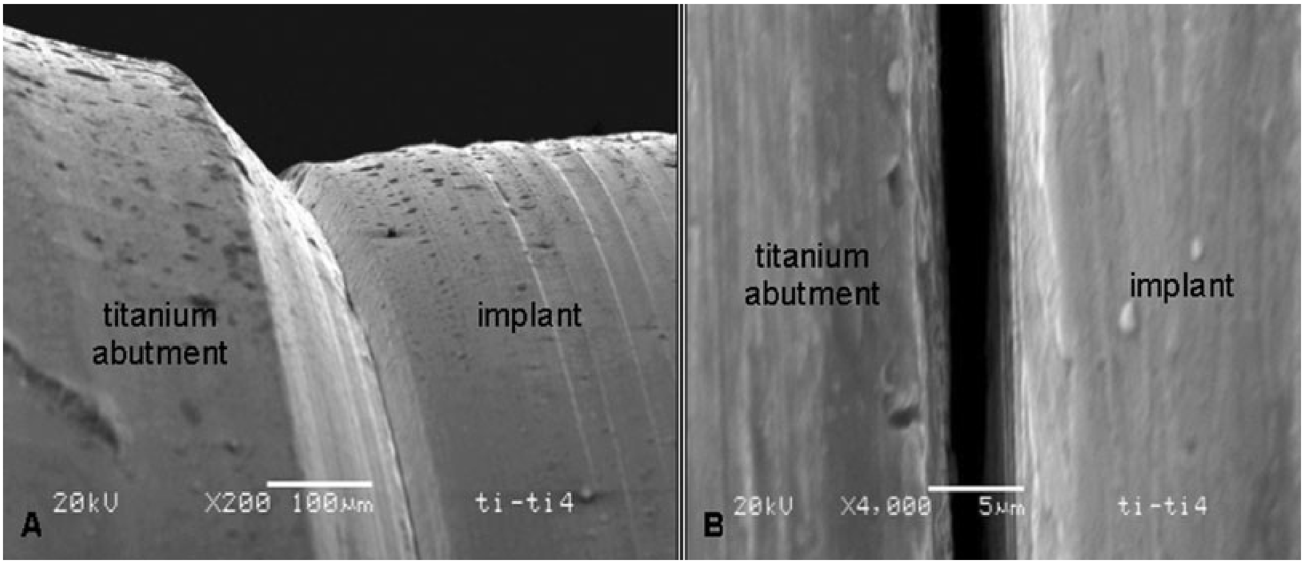

Figure 5 *A* and *B*, Scanning electron microscope photographs of titanium abutment group before the dynamic loading.

reasonable degree of protection against loosening.[14] In the present study, within-group comparisons after the dynamic loading revealed that microgap values were higher in the titanium abutment group in the palatinal site compared to the labial site. These results are comparable with the reports of the study by Merz and colleagues[14] and could be explained with the "back off" theory. As Cibirka and colleagues[15] stated, Bickford described the abutment screw as a spring stretched by preload that is maintained by the frictional fit of the threads. External forces can create a vibratory movement and cause the threads to "back off." The backing off of the threads leads to a reduction in the effective preload and diminishes the ability of the screw to maintain the joint stability thereby increasing the implant-abutment interface gap space. As a result, the mating surfaces of the titanium abutment and the implant are separated, which is also observed in the present study. Nevertheless, SEM evaluation of the screws or the interface after vertical sectioning of the implant/abutment complex was not undertaken in the present study and therefore, this study did not provide information

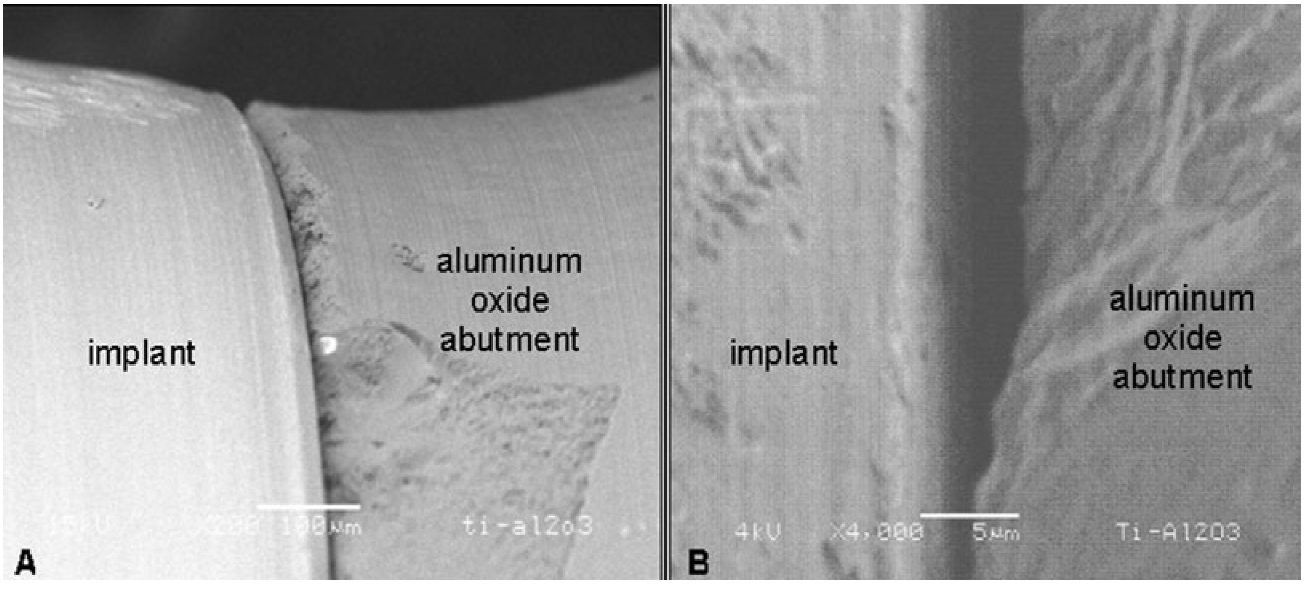

Figure 6 *A* and *B*, Scanning electron microscope photographs of aluminum oxide abutment group after the dynamic loading.

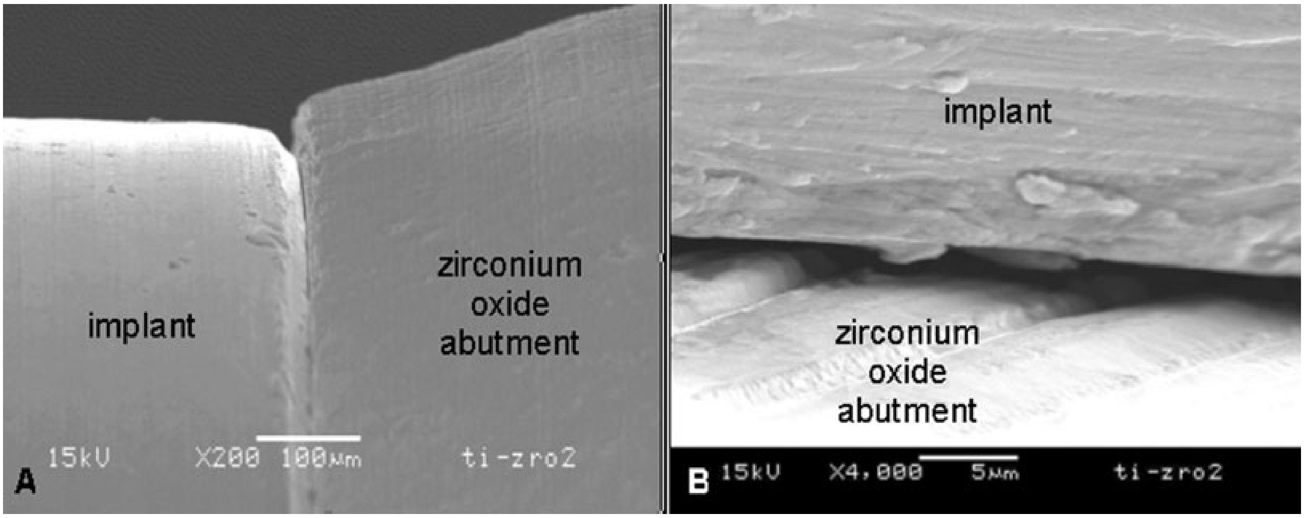

Figure 7 *A* and *B*, Scanning electron microscope photographs of zirconium oxide abutment group after the dynamic loading.

regarding possible wear or plastic deformation of the abutment screw.

In contrast with the findings of Abrahamsson and colleagues,[6] the results of the present study show that there were no significant changes seen among the abutment groups before and after loading except for the titanium abutment group. Abrahamsson and colleagues[6] stated that the greater bacterial contamination seen in ceramic abutments were because of the larger microgap formation compared to titanium abutments. When implant-abutment assembly was subjected to dynamic loading, micromovement may be seen at the implant-abutment interface. As a result of these movements, wear between the contacting surfaces takes place and these surfaces move closer.[22] When metal and ceramic are contact, the metal usually becomes abraded. Fretting wear occurs when repeated loading and unloading cause cyclic stresses that induce surface or subsurface breakup, resulting in the loss of material.[18] This could explain why there was no change between the abutment groups after

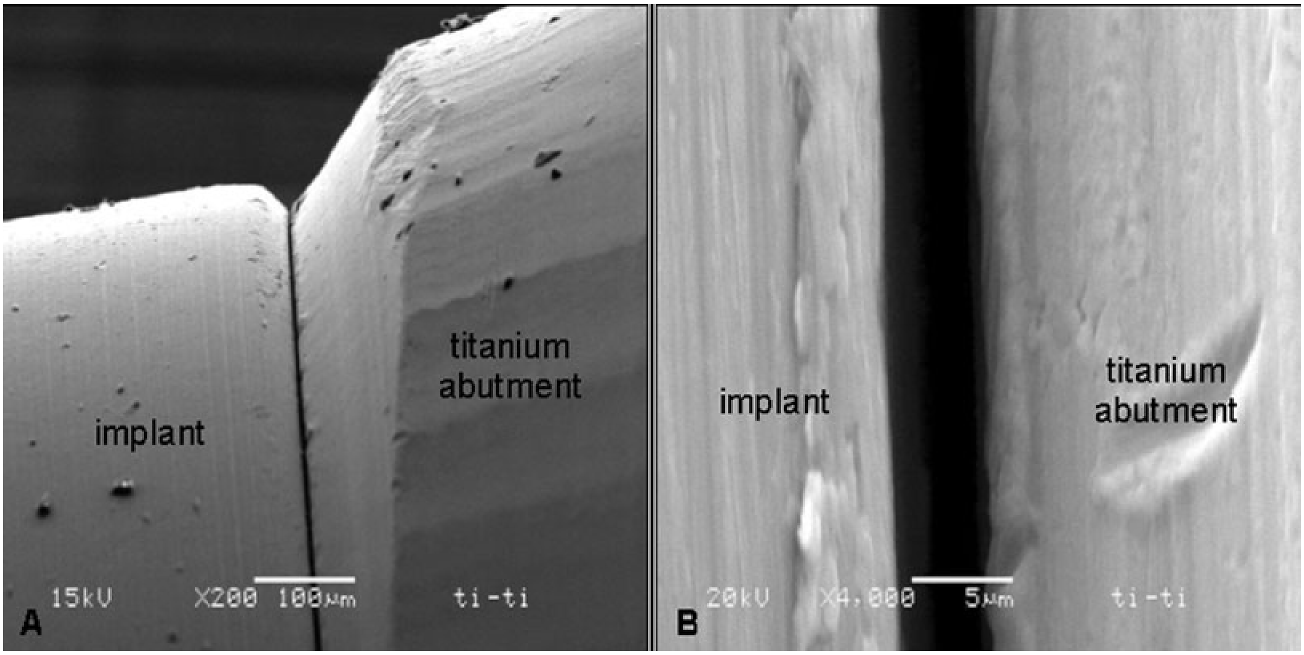

Figure 8 *A* and *B*, Scanning electron microscope photographs of titanium abutment group after the dynamic loading.

loading in the present study. The abutment screw may have backed off in all abutment groups. However, fretting wear occurring after loading in ceramic abutments may have caused the mating surfaces of ceramic abutments and implants to move closer instead of separating on the palatinal site. The reason why debris, an indicator of wear, was not evident at the implant-abutment interface in the SEM images depends on the fact that all the specimens were cleaned with ethyl alcohol in an ultrasonic cleaning bath for 5 minutes before SEM analysis prior to and after the experiments. Therefore, the implants and abutments, as well as the interface between those components, were thoroughly cleaned. Nevertheless, the loss of material at the outer edges of those components in the interface zone could, again, be well described by fretting wear, but in a different manner, as according to *Saint-Venant's principle*, the stresses remote from the point of application of the load are not affected by the precise behavior of the structure close to the point of application of the load. This implies that unlike the implant body or remote parts of the abutments, where no substance loss was observed in SEM, the outer edges of the abutments, which contact with implant neck during bending and dynamic loading, were subjected to high loads, leading to wear of those surfaces and loss of material.

A simulation of 500 mastication cycles was performed on an implant-supported restoration by Brodbeck[18], and the titanium debris abraded from the external hex by the aluminum oxide abutment was visible. Hoyer and colleagues[16] stated that implant/abutment joint opening under dynamic loading after 500,000 cycles was consistently in the range of 0 to 30 μm, and the joint opening was not significantly different as a function of cycle number. The amount of torque required to loosen the screws from three different antirotational screw-retained implant-abutment combinations was compared during simulated intraoral movements 1 and 6 months of loading. There was no significant difference in the torque necessary to loosen the screws for any of the implant systems when comparing the 6-month results to the 1-month results. It was stated that screw loosening is more likely to occur during the first month of function.[19] With these results as references, a 45-day simulation was used in the present study.

Although the specimen number is a limitation of the present study, these findings provide an initial step toward helping the clinician make a decision of the use of ceramic abutments for single-tooth implant restorations. Absences of abutment movement, coping, abutment, or screw fracture noted in any of the specimens after dynamic loading and a low number of loading cycles may have been a factor causing this observation in the present study design. Further research is needed to increase the understanding of the characteristics for success or failure of different types of ceramic abutments in clinical use today.

CONCLUSIONS

Within the limitations of this in vitro study, the following conclusions were drawn:

1. after the dynamic loading, there was no significant difference between the aluminum oxide, zirconium oxide, and titanium abutment groups regarding the microgap at the implant-abutment interface; and
2. for the palatinal surface comparisons, after the dynamic loading, the titanium abutment groups revealed significantly increased values when compared to the microgap measurements for the aluminum and zirconium oxide abutment groups.

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