Fracture resistance of titanium and zirconia abutments: An in vitro study

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Statement of problem. Little information comparing the fracture resistance of internal connection titanium and zirconia abutments exists to validate their use intraorally.

Purpose. The purpose of this study was to determine the fracture resistance of internal connection titanium and zirconia abutments by simulating cyclic masticatory loads in vitro.

Material and methods. Twenty-two specimens simulating implant-supported anterior single crowns were randomly divided into 2 equal test groups: Group T with titanium abutments and Group Z with zirconia abutments. Abutments were attached to dental implants mounted in acrylic resin, and computer-aided design/computer-aided manufacturing (CAD/CAM) crowns were fabricated. Masticatory function was simulated by using cyclic loading in a stepped fatigue loading protocol until failure. Failed specimens were then analyzed by using scanning electron microscopy (SEM) and fractographic analysis. The load (N) and the number of cycles at which fracture occurred were collected and statistically analyzed by using a 2-sample *t* test (α=.05).

Results. The titanium abutment group fractured at a mean (SD) load of 270 (56.7) N and a mean (SD) number of 81 935 (27 929) cycles. The zirconia abutment group fractured at a mean (SD) load of 140 (24.6) N and a mean (SD) number of 26 296 (9200) cycles. The differences between the groups were statistically significant for mean load and number of cycles (P<.001). For the titanium abutment specimens, multiple modes of failure occurred. The mode of failure of the zirconia abutments was fracture at the apical portion of the abutment without damage or plastic deformation of the abutment screw or implant.

Conclusions. Within the limitations of this in vitro study, 1-piece zirconia abutments exhibited a significantly lower fracture resistance than titanium abutments. The mode of failure is specific to the abutment material and design, with the zirconia abutment fracturing before the retentive abutment screw. (J Prosthet Dent 2013;109:304-312)

Clinical Implications

The fracture resistance of 1-piece zirconia abutments compared to titanium abutments under laboratory conditions suggests caution when prescribing regular-sized 1-piece zirconia abutments.

Restoring a dental implant in the esthetic zone can be challenging, especially if a metal implant abutment is directly visible or shows through the surrounding soft tissues. This is a common problem when implants are positioned too near the labial cortical bone plate or superficially in the alveolar bone[1] and may also be a problem in a patient with a thin gingival biotype or subsequent to crestal bone resorption around the dental

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implant. In addition, when esthetic demands justify the selection of ceramic crown materials, a low-value, metal abutment is difficult to mask. Implant abutments made of commercially pure titanium are well documented to be biocompatible[2,3] and have sufficient mechanical properties to support long-term fixed implant-supported dental prostheses.[2,4,5] However, when used in certain clinical situations, titanium abutments can create an unesthetic blue hue in the tissues[6-8] and may compromise the esthetic result if used in conjunction with ceramic crowns.

Several strategies are available to overcome these esthetic problems, including gold-colored titanium nitride-coated abutments and ceramic abutments made of alumina[9] or zirconia.[10] Zirconia has recently attracted significant interest because of its superior fracture resistance compared to alumina,[11] its superior esthetic properties,[6-8] and its improved biocompatibility compared to metal abutments.[12,13] Zirconia abutments can be broadly classified into 2 categories: 1-piece zirconia abutments, where the entire abutment is made of zirconia,[10] or 2-piece zirconia abutments consisting of a titanium or a titanium alloy element that engages with the dental implant and a transmucosal zirconia element.[8] A metal abutment screw is used to retain both types of zirconia abutment. Zirconia abutments may be manufactured as standardized components or customized by using computer-aided design/computer-aided manufacturing (CAD/CAM) technology.[14,15]

Several clinical studies have reported the short-term to medium-term survival of ceramic abutments with promising results.[9] However, zirconia may have a finite resistance to fracture,[16] and clinical reports of catastrophic fracture of zirconia abutments do exist, with significant biological and technical cost.[10]

Limited in vitro and clinical evidence that compares the mechanical properties of internal connection titanium abutments and newer, 1-piece, internal connection zirconia abutments is currently available. One limitation of several studies[10-13,17] is the use of static loading conditions, which is less clinically representative than applying cyclic loading regimens. Other in vitro studies[10-12,14] are limited because the specimens tested did not have a crown or dental implant as part of the assembly. Implant analog platforms are manufactured to replicate the dimensions of the corresponding implant platform but are usually made of stainless steel and differ in other physical dimensions. All these differences may affect the behavior of the implant-abutment complex. Kim et al[17] used static loading conditions to test 1-piece, CAD/CAM zirconia abutments attached to internal connection, regular platform, implant analogs and reported a mean (SD) fracture load of 480.01 (174.46) N. Mitsias et al[18] demonstrated in a pilot study (n=3) that the mean (SD) single load-to-failure value of zirconia abutments connected to 4.5 mm diameter internal connection implants was 690 (430) N.

Several in vitro studies[19-22] have used a combination of cyclic, thermal, and static loading protocols to test implant-abutment-crown assemblies. A low force (30 to 49 N) was applied to each specimen for 1.2 million cycles. The specimens were simultaneously subjected to thermal cycling between 5°C and 55°C for 60 second cycles. Although all zirconia specimens survived in the 4 studies, the clinical relevance of these low loads may be questioned as maximum occlusal forces have been demonstrated to be significantly higher than 30 to 49 N and have been reported to be as high as 370 N in the anterior region.[23] It is rare for abutments to fail at loads less than 50 N in laboratory conditions.[19-22] When specimens do not fail during testing, statistical analysis is difficult. Thermocyclic loading is relevant as ceramic materials suffer from slow crack growth in moist conditions. Aging and slow crack growth can be thermally activated and occurs in the presence of water or water vapor.[24]

As in vitro studies inevitably have limitations, any clinical interpretations should be expressed with caution. Many authors[10-13,17,19-22,25-27] have suggested that comparison can be made between the load-bearing capacities of specimens in the laboratory and expected loading conditions in the oral cavity. The survival of a representative sample of abutments subjected to a physiological range of loads in the laboratory has been thought to justify the clinical use of these abutments. However, several studies used static loads,[10-13,17] and the magnitude of load generated by masticatory activity and the contact angle of load are variable between individuals and may also vary at different times of the day.[28] Measurements of 40 to 370 N have been recorded between the incisor teeth[23,29-32] and 120 to 350 N between the canine teeth.[32] In addition, greater magnitude occlusal forces are expected in bruxers, and nocturnal parafunctional forces have been demonstrated to sometimes exceed the magnitude of maximum voluntary occlusal force.[28] Therefore, dental prostheses located in the anterior region of the jaws should be able to withstand a variable range of forces over an extended period of time in an aqueous environment. When selecting appropriate loads for laboratory testing, the researcher should consider the upper range of possible forces rather than the average loads and contact angles encountered in vivo.[33]

Well-designed, clinically relevant laboratory testing and data may indicate the likely clinical success of an implant-restorative option. Modification of the laboratory testing method can also be performed subsequent to clinical trials so that the types of failures seen are similar[34] before widespread clinical use is recommended. The null hypothesis was that the specimens containing zirconia abutments have the same fracture resistance as the specimens containing titanium abutments.

MATERIAL AND METHODS

Specimen Preparation

Twenty-two specimens were prepared for 2 test groups of 11 specimens each, representing implant-supported anterior single crowns (Fig. 1). Group T consisted of specimens with identical stock titanium abutments (TiDesign, 3.5/4.0, 4.5 mm diameter, 1.5 mm height; AstraTech Dental AB, Mölndal, Sweden), and Group Z consisted of specimens with identical 1-piece stock zirconia abutments (ZirDesign 3.5/4.0, 4.5 mm diameter, 1.5 mm height; AstraTech Dental AB). Twenty-two identical dental implants (OsseoSpeed; AstraTech Dental AB) 4.0 mm in diameter and 9.0 mm in length were held in position with an impression coping (Implant Pick-up 3.5/4.0; AstraTech Dental AB) attached to a drill press (BF 1; Bredent GmbH, Senden, Germany), which acted as a device to standardize the mounting position. The implants were then mounted centrally and parallel to a sectioned polyvinyl chloride (PVC) pipe (25 mm diameter; 18 mm high) and embedded in an autopolymerizing acrylic resin (Unifast Trad III; GC Corp, Tokyo, Japan) to a height just below the implant collar.[21,22,35] The acrylic resin was allowed to completely polymerize over 24 hours.

The abutments were numbered and randomly assigned to an implant by using a random-number generator (Microsoft Excel 2007; Microsoft Corporation, Redmond, Wash). The abutments were connected to their allocated implant via a titanium abutment screw tightened to a torque value of 20 Ncm with a calibrated torque wrench (AstraTech Dental AB). To minimize variability in the abutment size and thickness, and to eliminate the weakening effect of preparing the abutment, the crowns were designed such that adjustment of the stock abutments was not necessary. This also prevented the possible phase-transformation caused by grinding the zirconia abutment.[36] A diagnostic waxing of a maxillary right central incisor crown based on the anatomic average[37] was made to encompass an unprepared stock abutment. Based on the wax pattern, identical CAD/CAM base metal crowns (Coron; Insitute Straumann AB, Gothenburg, Sweden) were copy-milled from blanks by using the Etkon system (Insitute Straumann AB). To ensure accurate seating and marginal adaptation, the crowns were hand finished with a tungsten carbide bur (Man-tc-1559; Mani Inc, Tochigi, Japan) under stereo microscopic evaluation at ×10 magnification (G20XT; Tokyo Kinzoku Co, Ltd, Tokyo, Japan). The dimensions of the crowns were 8.6 mm at the widest mesiodistal portion of the tooth, 13.0 mm at the longest apicocoronal distance (CEJ to incisal point), and 6.3 mm at the widest buccolingual distance.

1 Implant components (top to bottom), including base metal alloy crown, titanium screw, titanium and zirconia abutments, and 4-mm-diameter internal connection implant.

Polytetrafluoroethylene tape was placed over the abutment screw, and the crowns were cemented to their allocated abutment with a commercially available resin cement (Panavia F2.0; Kuraray Dental, Tokyo, Japan). The specimens were then stored in saline (sodium chloride 0.9%; Baxter Healthcare Pty, Ltd, Old Toongabbie, Australia) for 24 hours at room temperature. Specimens were standardized except for the abutment material, which differed between the test groups.

Fatigue testing

Masticatory forces were simulated by using closed-loop servohydraulics (MTS 810 Materials Test System; MTS Systems Corp, Eden Prairie, Minn) (Figs. 2, 3). Each specimen was placed in a customized brass device so that the long axis of the crown was at an angulation of 30 degrees to the loading platen of the servohydraulic testing machine to simulate a Class I incisor relationship.[38] The rounded, metal loading platen was positioned on the palatal surface of the crown, 2 mm from the incisal edge. The masticatory cycle was simulated by an isometric contraction (load control) applied through the metal loading platen. Graphite (Dixon's Microfyne Graphite; Thomas Grozier & Son, Lane Cove, Australia) was used as a lubricant between the platen and the crown. The specimens were kept moist during testing by using gauze soaked in saline to cover the specimens.

2 Assembled specimen fastened to adjustable (X-Y plane) customized device attached to servohydraulic machine.

3 Assembled specimen being cyclically loaded at 30 degrees to long axis.

A pilot study with 2 titanium and 2 zirconia abutment specimens similar to those in the main study was conducted to determine the single load-to-failure values. With the data from the pilot study, statistical software (Sample Power 20; IBM Corp, Armonk, NY) was used to determine the appropriate number of specimens required to provide statistical significance at a power of .80.

Twenty-two specimens were then prepared and cyclically loaded in a stepped fatigue loading protocol[39-41] at a frequency varying between 120 and 300 masticatory cycles per minute (2 to 5 Hz) depending on the maximum load being applied (load control).[40-46] The servohydraulic machine was programmed to provide a consistent load during fatigue testing rather than consistent displacement of the specimen to account for flexure. Cyclic load was applied starting at an initial load of 50 N for 5000 cycles (preconditioning phase). This was followed by stages of 100, 150, 200, 250, 300, and 400 N at a maximum of 20 000 cycles each.[39-41] Specimens were loaded until fracture. Failure was determined by an audible crack or by automatic software detection (MTS 810 Materials Test System; MTS Systems Corp). A sudden increase in displacement of the specimen away from the loading platen or a sudden reduction in force applied to the specimen automatically triggered a machine interlock stop. The number of endured cycles and the maximum load applied at failure were recorded. The data were statistically analyzed by a 2-sample *t* test (α=.05) by using statistical software (Minitab, v16.1.1; Minitab Inc, State College, Pa). The outcomes seemed to satisfy the assumption of normality but not homogeneity of variances, so *t* tests with unequal variances were used.

Failed specimens were analyzed to determine the mode of failure and identified as abutment fractures, abutment screw fractures, abutment deformation, abutment screw deformation, or implant deformation, and a specimen could have multiple failure modes. Specimens were further analyzed with a stereomicroscope (Leica S8APO; Leica Microsystems GmbH, Wetzlar, Germany) and Scanning Electron Microscopy (SEM) (Quanta Scanning Electron Microscope; FEI Company, Hillsboro, Ore) to identify fracture location and perform fractographic analysis.[47]

RESULTS

The results of the study are shown in Table I. The titanium abutment group fractured at a mean (SD) load of 269.6 (56.7) N and a mean (SD) of 81 935 (27 929) cycles. The zirconia abutment group fractured at a mean (SD) load of 139.8 (24.6) N and a mean (SD) of 26 296 (9200) cycles. The difference was statistically significant for both mean load and mean number of cycles (P<.001). The survival rate of titanium abutments was significantly higher than that of zirconia abutments (P<.001).

The mode of failure for the 2 groups is reported in Table II. For the titanium abutment specimens, multiple modes of failure occurred, including fracture or plastic deformation of the abutment screw and plastic deformation of the abutment and the implant. The mode of failure of the zirconia abutments was fracture at the apical portion of the abutment without damage or plastic deforma-

TABLE I. Number of cycles before failure and maximum load before failure

Titanium Abutments			Zirconia Abutments		
Specimen No.	Total Cycles Before Failure	Max Load Before Failure (N)	Specimen No.	Total Cycles Before Failure	Max Load Before Failure (N)
1	66 808	230.0	2	46 664	188.0*
4	25 850	160.0*	3	27 200	164.0*
5	46 291	200.0	7	26 061	147.5*
6	73 213	250.0	10	5568	89.3*
8	113 086	361.0	12	26 029	140.0*
9	88 118	284.0*	14	26 026	130.0*
11	108 030	306.0	17	26 087	151.0*
13	89 400	270.0*	18	25 950	125.0*
15	102 287	290.0*	19	26 916	131.0*
16	94 354	300.0	21	26 020	136.2*
20	93 848	315.0	22	26 734	135.3*
Mean	81 935	269.6		26 296	139.8
SD	27 929	56.7		9200	24.6

* Specimen fractured while increasing load.

TABLE II. Location of failure of specimens

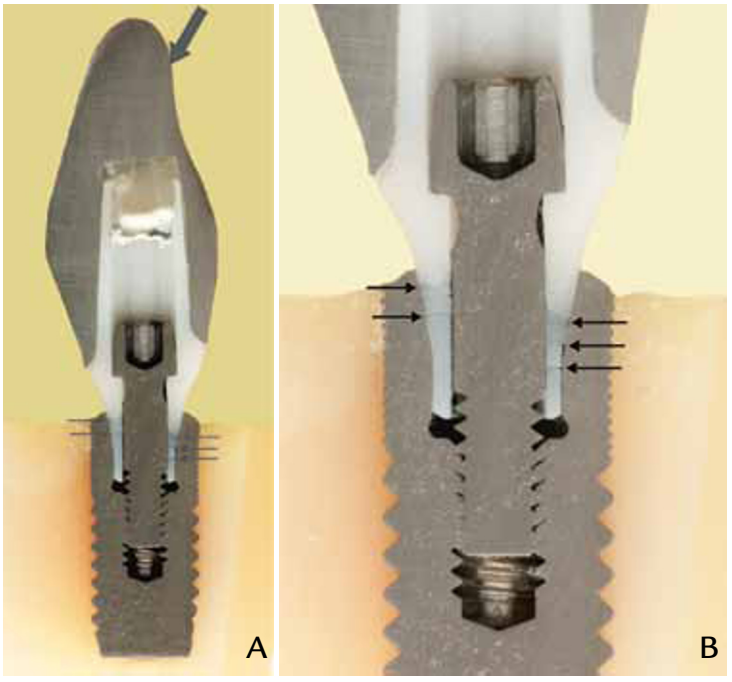
		Location of Failure				
Group	Total Number of Abutments	Abutment Fracture	Abutment Screw Fracture	Abutment Deformation	Abutment Screw Deformation	Implant Deformation
Titanium	11	0	10	11	11	11
Zirconia	11	11	0	0	2	1

4 Typical fracture of zirconia abutment propagating from palatal side of hexagon (thinnest portion of abutment).

tion of the abutment screw or implant (Figs. 4, 5). All zirconia specimens fractured at the internal hexagon portion, which was the thinnest section of the abutment.

DISCUSSION

The null hypothesis was rejected as the zirconia abutments demonstrated a significantly lower fracture resistance than the titanium abutments. This study used a stepped fatigue loading protocol,[39-41] in which a predetermined load was applied for a defined number of cycles, followed by incremental increases in load for a set number of cycles until failure



5 A, Cross-sectional view showing load application point on crown (large arrow). Multiple fractures in apical portion of abutment (small arrows). B, Fractures in apical portion of abutment (arrows).

of the specimen. The benefit of this type of test is that it provides a better simulation of clinical conditions than a static load test,[39-41] and does not require an extensive period of testing. Several in vitro studies[10-13,17,25] investigating zirconia abutments have shown that high magnitude static loads are required to fracture specimens. Cyclic loading has been demonstrated to decrease the fracture resistance of zirconia abutments. Gehrke et al[26] reported a decrease in the strength of zirconia abutments from 672 N to 405 N after cyclic loading. Other studies[19-22] have applied low-magnitude cyclic loads without any resultant failures and have required subsequent static loading to provide statistically significant data. However, the clinical relevance of these tests may be questioned.[18,39-41]

Mitsias et al[18] used a similar experimental design to the present study and tested specimens in a step-stress accelerated testing protocol. The titanium (Profile BiAbutment 4.5/5.0; AstraTech AB) and zirconia abutments (Ceramic Abutment 4.5/5.0; AstraTech AB) tested were designed for 4.5 mm diameter implants. The calculated reliability for the titanium abutment group at 50 000 cycles and a load of 400 N was 1.00 (2-sided 90% confidence bound: 1.0-0.93). The zirconia abutment group at 50 000 cycles and a 175-N load had a reliability of 0.83 (2-sided 90% confidence bound: 0.96-0.42).[18] It would be expected that smaller diameter abutments with thinner walls would have a lower fracture resistance.

Accelerated testing is sometimes desirable for in vitro experiments before the significant commitment of time and expense required for a clinical trial is made.[42-44] Certain studies have increased the physiologic mean of 70 masticatory cycles/minute (1.17 Hz)[45] to 5 Hz.[39-41] However, increasing the cycling speed above this value may require correction factors[46] and increases in cyclic loading speed should be moderate to gain clinically relevant laboratory data.

This study examines bone-level, internal-connection implants, which may be used in esthetically critical areas of the mouth. The implant size used in this study was the 4.0 mm-diameter internal connection implant rather than the larger 4.5-mm diameter implant used in other studies.[18,21] In clinical practice, anatomic constraints in the anterior region of the jaw may often limit the diameter of the implant to 4.0 mm.

The mode of failure of the zirconia abutments was fracture at the apical portion of the abutment without damage or plastic deformation of the abutment screw or implant and was consistent with results reported by Mitsias et al[18] and Nothdruft et al.[21] All specimens in the titanium abutment group showed a degree of deformation of the metallic components. Titanium allows some favorable degree of elastic deformation during screw tightening and accommodates the plastic deformation generated by friction between the different components.[4] This is known as the settling effect.[5] As loads increase beyond the yield limit of the titanium abutment, the components deform and bend, which may lead to the eventual fracture of the weakest component, the abutment screw, particularly after cyclic loading.

Dental prostheses located in the anterior region of the jaw should be able to withstand a variable range of forces over an extended period of time in an aqueous environment.[24] When selecting appropriate loads for in vitro testing, the upper limit of the possible loads encountered[33] should be considered rather than the mean loads found in vivo. Simultaneous thermocyclic loading of specimens with zirconia abutments was not practical for this present study but may result in a decreased number of cycles to failure and a lesser mean maximum applied force before failure.

Although in vitro studies should be as clinically relevant as possible[34] and use standardized specimens,[39] the present study was designed to limit the variables solely to the abutment materials. Therefore, the crowns were made of a cobalt chrome alloy instead of the ceramic materials, which would be commonly placed clinically. The use of ceramic crowns may have introduced other materials and interfaces at which failure may have occurred, but this was not the focus of the present

study. The dimensions of the crowns were also designed to avoid preparation of the abutments and negate the possible detrimental effect of grinding the abutments.[36] However, the crown dimensions were still within the physiologic range.[37] Magne et al[37] demonstrated that the maximum dimensions of a central incisor crown in white participants was 11.07 mm at the widest mesiodistal portion of the tooth and 13.51 mm at the longest apicocoronal distance (CEJ to incisal point). The greater the length of the crown, the greater the lever arm force that can be applied to the abutment-implant interface. All abutments had an anatomical marginal configuration, which provided antirotational resistance and precluded the need for preparing mechanical antirotational features.

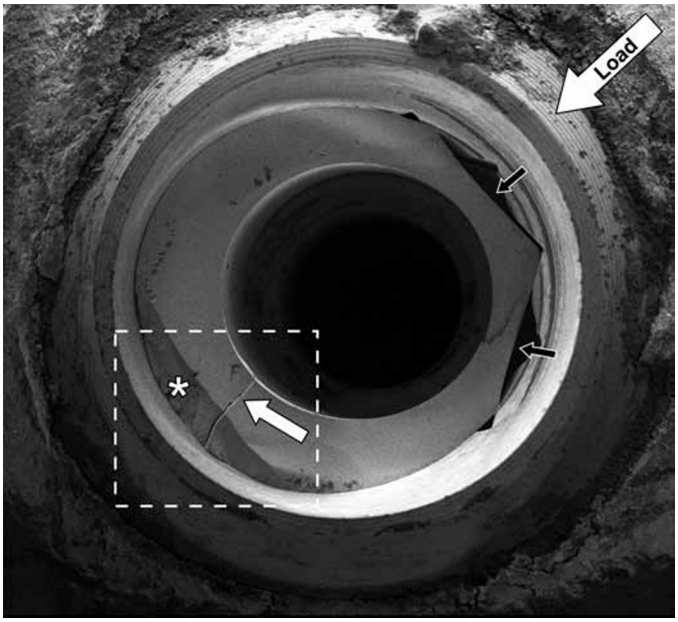
In this current study, implants were embedded in acrylic resin supported by a PVC ring. This technique of mounting implants or implant analogs in autopolymerizing acrylic resin is consistent with several in vitro studies.[20,21,35] It may be beneficial to use a material that has a modulus of elasticity and a shape and volume more closely matched to alveolar bone in the anterior maxilla as this may have a better stress-distribution effect. In vitro studies are also generally unable to accurately reproduce dynamic occlusal movements and patterns. In this study, a contact angle of 30 degrees was chosen to represent an interincisal angle of 150 degrees in a Class I occlusion.[38] Other studies have used contact angles of 30 to 60 degrees,[10,11,13,26] and a contact angle of 30 degrees is recommended by Food and Drug Administration guidelines.[33] It has been suggested that for single implant prostheses, laterotrusive contacts should be distributed to the natural dentition rather than to the prosthesis.[21] However, not all implant prostheses are positioned in favorable occlusal relationships, and clinical judgment may be required to select the appropriate abutment material in these situations. In this study, specimens in both experimental groups failed at loads considered to be within the physiologic range. This may possibly be explained by the loading of isolated, single specimens with a significant lever arm rather than the clinical scenario in which the prosthesis is often protected by a natural dentition. Simulating dynamic occlusal patterns to test implant abutments is technically possible, but economic and time limitations are usually prohibitive and such a complex arrangement may be unnecessary.

6 Clinical image of failed zirconia abutment in service for 18 months. Courtesy Dr A. Dillon.

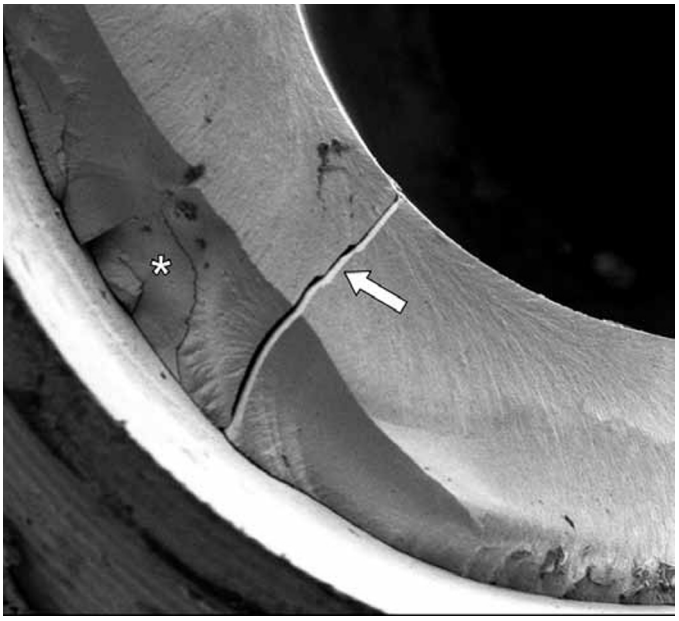
Zirconia abutments have 2 primary design variations: the 1-piece and 2-piece design. In vitro results showed a median (SD) fracture resistance of 294 (53) N for external hexagon, 4.0 mm collar, 2-piece, titanium-reinforced abutments after cyclic loading for 1.2 million cycles and static loading until failure compared to titanium abutments, which had a median (SD) fracture resistance of 324 (85) N under the same conditions.[20] Further research should be undertaken to compare the fracture resistance of 2-piece zirconia abutments with both titanium and single-piece zirconia abutments.

Both 1-piece and 2-piece designs can be custom-manufactured by using CAD/CAM technology. The perceived advantage of CAD/CAM is that the zirconia is milled in its green or soft state and then sintered in its final shape with the maximum amount of tetragonal phase (Y-TZP).[14,15] Further adjustment performed after sintering will induce the transformation from the tetragonal to monoclinic phase that will initially increase the strength of zirconia as it generates compressive stresses. This crack-resisting phenomenon is finite, and the material may eventually fracture after repeated loading once the transformation toughening effect is overcome.[16] Clinicians should be aware that postsintering adjustment is often necessary.

In this in vitro study, failure of the zirconia abutments occurred at the apical hexagon, the thinnest portion of the abutment. Upon cyclic loading, plastic deformation of the screw thread and subsequent deflection of the whole complex is assumed to occur. Stress and tension develop within the thinnest ceramic portion, and fatigue-assisted crack initiation and growth lead to fracture. In the present study, analysis of the mode of failure of zirconia abutments is consistent with some clinical failures (Fig. 6). SEM analysis of the failed laboratory specimens revealed compression curls



7 Representative scanning electron microscope image (×29) demonstrating fracture in region of hexagon (small arrows) and opposing bend fracture characterized by compression curl (asterisk) and crack (large arrow). Direction of load application is indicated (large arrow).



8 Higher magnification (×111) scanning electron microscope image demonstrating compression curl (asterisk) and crack (arrow) propagating through thickness of abutment wall.

(Fig. 7) that are indicative of bend fractures, which usually occur opposite the crack initiation site.[47] A higher magnification SEM (Fig. 8) also revealed cracks that extended through the thickness of the abutment wall. It is assumed that some secondary fractures occurred more coronally as the abutment deflected within the implant body.

CONCLUSIONS

1-piece titanium and zirconia abutments were tested in a stepped fatigue loading protocol. Within the limitations of this in vitro study, the titanium abutment system was significantly more fracture resistant than the zirconia abutment system. The following conclusions can be made:

1. The mean number of cycles until failure of the titanium abutment group was 3 times that of the zirconia abutment group.

2. The average load before failure for the titanium abutment group was almost twice that of the zirconia abutment group.

3. Specimens in both experimental groups failed at loads considered to be within the physiologic range.

4. Caution must be exercised when prescribing regular-sized, single-piece zirconia abutments, and they should only be considered in low occlusal load situations where esthetics are paramount.

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