Stability of Prototype Two-Piece Zirconia and Titanium Implants after Artificial Aging: An In Vitro Pilot Study

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ABSTRACT

Background: Zirconia oral implants are a new topic in implant dentistry. So far, no data are available on the biomechanical behavior of two-piece zirconia implants. Therefore, the purpose of this pilot investigation was to test in vitro the fracture strength of two-piece cylindrical zirconia implants after aging in a chewing simulator.

Materials and Methods: This laboratory in vitro investigation comprised three different treatment groups. Each group consisted of 16 specimens. In group 1, two-piece zirconia implants were restored with zirconia crowns (zirconia copings veneered with Triceram®; Esprident, Ispringen, Germany), and in group 2 zirconia implants received Empress® 2 single crowns (Ivoclar Vivadent AG, Schaan, Liechtenstein). The implants, including the abutments, in the two zirconia groups were identical. In group 3, similar titanium implants were reconstructed with porcelain-fused-to-metal crowns. Eight samples of each group were submitted to artificial aging with a long-term load test in the artificial mouth (chewing simulator). Subsequently, all not artificially aged samples and all artificially aged samples that survived the long-term loading of each group were submitted to a fracture strength test in a universal testing machine.

For the pairwise comparisons in the different test groups with or without artificial loading and between the different groups at a given artificial loading condition, the Wilcoxon rank-sum test for independent samples was used. The significance level was set at 5%.

Results: One sample of group 1 (veneer fracture), none of group 2, and six samples of group 3 (implant abutment screw fractures) failed while exposed to the artificial mouth. The values for the fracture strength after artificial loading with 1.2 million cycles for group 1 were between 45 and 377 N (mean: 275.7 N), in group 2 between 240 and 314 N (mean: 280.7 N), and in the titanium group between 45 and 582 N (mean: 165.7 N). The fracture strength results without artificial load for group 1 amounted to between 270 and 393 N (mean: 325.1 N), for group 2 between 235 and 321 N (mean: 281.8 N), and between 474 and 765 N (mean: 595.2 N) for the titanium group. The failure mode during the fracture testing in the zirconia implant groups was a fracture of the implant head and a bending/fracture of the abutment screw in the titanium group.

Conclusions: Within the limits of this pilot investigation, the biomechanical stability of all tested prototype implant groups seems to be – compared with the possibly exerted occlusal forces – borderline for clinical use. A high number of failures occurred already during the artificial loading in the titanium group at the abutment screw level. The zirconia implant groups showed irreparable implant head fractures at relatively low fracture loads. Therefore, the clinical use of the presented prototype implants has to be questioned.

KEY WORDS: all-ceramic crowns, artificial mouth, fracture strength, two-piece zirconia implant

INTRODUCTION

Zirconia oral implants have been promoted, at least on the European market, for some years. Most of the systems are of a one-piece screw design. However, there

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DOI 10.1111/j.1708-8208.2008.00116.x

are ambitions to fabricate two-piece zirconia implants comparable with the two-piece titanium implants. Irrespective of implant design, all zirconia implant systems lack scientific background. Zirconia was applied relatively early as an oral implant coating material in animal investigations.[1,2] Animal investigations showed that the bone integration capacity of zirconia seems not to be different from that of titanium.[3–8] Zirconia ceramic is biocompatible[4,9] and less prone to plaque accumulation compared with metal substrates.[10–12]

So far, there is only minimal information regarding the biomechanical behavior of zirconia oral implants. Only two in vitro investigations evaluated the physical properties of zirconia implants in comparison with similar-shaped titanium implants in a finite element analysis[13] and in a long-term loading/fracture testing investigation.[14] In the former investigation, three-dimensional finite element analysis models of a maxillary incisor with ReImplant® implants (ReImplant, Hagen, Germany)[15] were made, surrounded by cortical and cancellous bone. A porcelain-fused-to-metal (PFM) crown for a titanium implant and a ceramic crown for a zirconia implant were modeled and the stress levels were calculated according to the von Mises criteria. Zirconia and titanium implants showed similar stress distributions. In the latter in vitro investigation on the biomechanical behavior, two groups of two-piece zirconia implants with the shape of a central incisor (ad modum ReImplant) were tested in an artificial mouth. The implants seemed to possibly fulfill the biomechanical requirements for anterior teeth showing mean fracture loads of the crowns of about 550 N after loading for 1.2 million cycles. The peculiarity of the two mentioned in vitro investigations was that the implants used had the shape of a tooth root. As this design of root analogue implants could not be successfully used clinically,[16] a cylindrical implant form was generated for the zirconia implants.

The present investigation is a successive in vitro investigation of the investigation by Kohal and colleagues.[14] As there are no data available on the fracture strength and, therefore, on the safety of cylindrical, two-piece zirconia implants, the primary objective of the present investigation was to evaluate the fracture strength of such prototype zirconia implants, which were restored with different all-ceramic crowns using a universal fracture machine. Also, the influence of artificial loading over a period that mimics 5 years of clinical service was investigated. The secondary objective of the present investigation was whether different ceramics used for crown fabrication (lithiumdisilicate vs zirconia) influence the stability of the zirconia implants. A prototype titanium implant system – similar to the zirconia implants – restored with PFM crowns served as the control group.

MATERIALS AND METHODS

For this in vitro investigation, 16 prototype cylindrical titanium (commercially pure titanium grade 2) implants (Department of Prosthodontics, University Clinic Freiburg, Freiburg, Germany) were fabricated using the ReImplant system (Table 1).[17] On screw-retained prefabricated titanium abutments (Department of Prosthodontics, University Clinic Freiburg), PFM crowns were cemented (Ketac™ Cem; 3M ESPE, Seefeld, Germany) (Figure 1). The PFM crowns for the titanium implant group were produced using a noble gold alloy (SMK 84; Wieland, Pforzheim, Germany) and Vita Omega 900 ceramic (Vita Zahnfabrik, Bad Säckingen, Germany) as veneering material.

Thirty-two two-piece prototype cylindrical zirconia implants were fabricated out of yttria-stabilized tetragonal zirconium dioxide polycrystal (Department of Prosthodontics, University Clinic Freiburg). Prefabricated zirconia abutments with a conical base that fitted into the implants were cemented into the implant head after airborne particle abrasion (Al_2O_3 powder: 50 µm; air

TABLE 1 Distribution of Test and Control Groups

Group	Chewing Simulator (*n*)	Without Chewing Simulator (*n*)	Fracture Load (with/without chewing simulator)
ZrO_2 + DCS/Triceram (group 1)	8	8	7/8
ZrO_2 + Empress 2 (group 2)	8	8	8/8
Titanium + PFM (group 3)	8	8	2/8

Titanium = titanium implant; PFM = porcelain-fused-to-metal crown; ZrO_2 = zirconia implant; DCS/Triceram = DC-Zirkon coping veneered with Triceram.

Figure 1 Titanium implant with abutment (*right*), zirconia implant with abutment (*left*).

pressure: 2 bars) with Panavia® 21 (Kuraray, Tokyo, Japan) (see Figure 1). Sixteen zirconia implants were restored with zirconia copings, which were produced using the Precident® system (DCS-Dental, Allschwil, Switzerland). The frameworks had a uniform thickness of 0.6 mm and were veneered with Triceram® ceramic veneering material (Espident, Ispringen, Germany; group 1). The other 16 zirconia implants received Empress® 2 crowns (Ivoclar Vivadent AG, Schaan, Liechtenstein; group 2). For the Empress group, Empress 2 single-crown frameworks with a homogenous thickness of 0.6 mm were fabricated. The frameworks were then veneered with Eris® (Ivoclar Vivadent) for Empress.

In order to fabricate similar crowns for each group, a "master crown" was produced and indexed with silicone. According to the "master crown," all PFM, DCS/Triceram, and Empress 2 crowns were fabricated. Thus, a relatively homogenous group of single crowns with regard to thickness and dimensions were obtained.

Preparatory steps for crown cementation included the etching of the intaglio surface of the Empress 2 crowns with hydrofluoric acid. The intaglio surface of the PFM and of the zirconia crowns were airborne-particle abraded with 50-µm Al_2O_3-powder at a pressure of 2 bars. The crowns were cleaned finally in an ultrasonic bath. Furthermore, the implant heads/shoulders and abutments were cleaned with a bicarbonate airflow device (Plaquesweep®; Neubauer Dental, Offenbach, Germany), then rinsed with water, dried, cleaned with 70% isopropanol, and air dried. The PFM crowns were cemented using glass ionomer cement (Ketac Cem). The two all-ceramic crown groups were adhesively cemented using Panavia 21. Before cementation, the intaglio surfaces of the Empress 2 and zirconia crowns was treated using a silane coupling agent (Clearfil SE Bond Primer and Clearfil Porcelain Bond Activator; Kuraray). Cement excess was removed and where Panavia was used, an air block (Oxyguard®; Kuraray) was applied.

Eight samples of each experimental group were subjected to be loaded in a chewing simulator (Willytec, Munich, Germany) (Figure 2). The samples were therefore placed into sample holders and stabilized with autopolymerizing acrylic resin (Technovit® 4000; Heraeus Kulzer GmbH & Co., Wehrheim, Germany) at an angle of approximately 135 degrees to the horizontal plane. As antagonistic material, steatite ceramic balls with a diameter of 6 mm were used.

Testing in the Chewing Simulator (Artificial Mouth)

Eight samples of each experimental group were loaded for 1.2 million chewing cycles in the computer-controlled dual-axis chewing simulator. This procedure mimicked 5 years of clinical functional loading.[18] The load that was applied amounted to 45 N and was placed approximately 2 mm below the incisal edge of the crowns. Furthermore, the samples were exposed to a computer-controlled thermal load using thermocycling (10,000 cycles, 5°C and 55°C for 60 seconds each, with a pause between cold and warm water of 12 seconds). For recording any events (ie, chipping of the veneering material, fractures of the crowns or implants), all samples were examined twice a day. After artificial aging in the simulator, the crowns were evaluated under a stereomicroscope (Axioskop; Zeiss, Jena, Germany) at a magnification of ×25 for possible crack initiation and propagation.

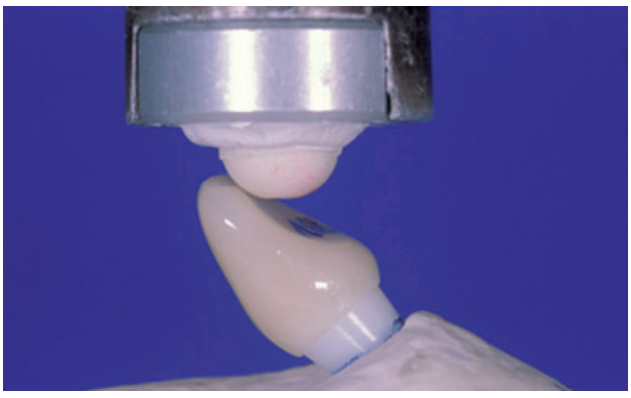

Figure 2 Implant/crown system subjected to artificial load.

Testing of the Fracture Strength in a Universal Testing Machine

All artificially aged (eight samples) and also the non-aged specimens of each group (eight samples) were tested regarding fracture strength in a universal testing machine (model 1445; Zwick, Ulm, Germany). A load was applied onto the implant crowns under a crosshead speed of 1.5 mm/min at an angle of 135 degrees to the horizontal plane. The loads required for fracturing the samples were recorded on an X-Y writer (Spare 2000; Kipp & Zonen, Delft, Netherlands). Failure was recorded when a first sharp drop down of the graphical curve could be depicted (fracture of the ceramic, bending of the abutments).

STATISTICAL ANALYSIS

A fracture load of 45 N was assigned to all specimens that fractured during the artificial loading process. For the pairwise comparisons in the different test groups with or without artificial loading and between the different groups at a given artificial loading condition (PFM vs Empress 2, PFM vs DCS/Triceram, Empress 2 vs DCS/Triceram), the Wilcoxon rank-sum test for independent samples was used. The significance level was set at 5% (Statistic Software "R"; R Development Core Team, R Foundation for Statistical Computing, Vienna, Austria).

RESULTS

Results after the Chewing Simulation Test

One specimen in the DCS/Triceram group (group 1) showed a fracture of the veneering material at 340,000 chewing cycles. All other specimens in this group survived the 1.2 million cycles in the artificial mouth without visible damage. No failure was observed in the Empress 2 group (group 2). However, in the titanium implant group, six adverse events occurred: one specimen showed a fracture of the abutment screw at 100,000 cycles. Between 600,000 and 1,000,000 cycles, five additional abutment screws fractured. Only two of the previous eight PFM specimens survived the artificial loading.

Results of the Fracture Strength Test

When analyzing the mode of fracture in the all-ceramic groups (DCS/Triceram crowns on zirconia implants, Empress 2 crowns on zirconia implants) in 20 (14 in the DCS/Triceram group, 6 in the Empress 2 group) of the cases, a fracture of the buccal aspect of the implant head occurred (Figure 3). In addition, in 11 samples (1 in the DCS/Triceram group, 10 in the Empress 2 group), there was a fracture line through the middle of the crowns with a palatal and labial fracture segment. This fracture line continued into the implant heads. In the PFM group, only two of the eight samples from the chewing simulation could be fracture loaded. In this group, a bending of the abutment screw occurred, with a chipping at the buccocervical region of the veneering material.

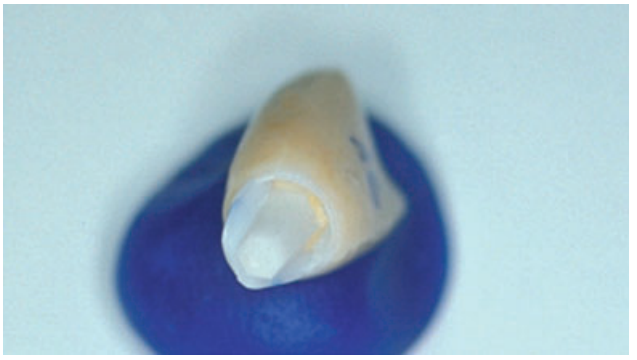

Figure 3 Fracture of the implant unit at the implant head.

The results of the fracture strength tests are presented in Table 2. As there were no significant differences in fracture values between the two all-ceramic restorations groups, they were combined. The values for the titanium implants with the PFM crowns without artificial load varied between 473 and 764 N. For the all-ceramic group without artificial loading, the fracture

TABLE 2 Mean and Median Values and SDs of the Fracture Strength Test of PFM and All-Ceramic Crowns (in N)

	AllC – C.	AllC + C.	PFM – C.	PFM + C.
Mean	302.9	277.6	595.2	165.7
SD	43.0	77.7	102	251.4
Median	$305^{A,2}$	$280.5^{a,2}$	$588.8^{B,1}$	$45^{a,1}$

Superscript uppercase letters indicate the different groups without chewing load; identical superscript uppercase letters indicate absence of significant difference. Superscript lowercase letters indicate the different groups with chewing load; identical superscript lowercase letters indicate absence of significant difference. Superscript numbers indicate same group with or without chewing load; identical numbers indicate absence of significant difference between chewing load and no chewing load.
PFM = porcelain-fused-to-metal crown; AllC = mean fracture strength values of all all-ceramic crowns (Empress 2 and Zirconia combined); – C. = without chewing simulation; + C. = with chewing simulation.

load ranged from 235 to 392 N. With artificial loading, the values for the titanium implants with the PFM crowns were between 45 and 582 N and between 45 and 377 N for the all-ceramic group.

There was no statistically significant difference in any of the groups with regard to the fracture test values when artificial load was compared with no artificial loading (titanium and PFM: $p = .091$; zirconia and all-ceramic crowns: $p = .1370$). When not artificially loaded, the fracture test values among the different groups were significantly different. The PFM group showed significantly higher fracture values compared with the all-ceramic ($p = .0143$) group. There was, however, no statistically significant difference between the different groups when artificially loaded ($p = .2720$).

DISCUSSION

This report is the first investigation to evaluate in vitro two-piece cylindrical zirconia implants restored with all-ceramic crowns. All implants used were prototype implants and are not yet available on the market. The high occurrence of failures in the titanium/PFM group during artificial loading excludes this implant system from clinical use at the moment. The weak link in the titanium/PFM group was the abutment screw, which broke during artificial loading and which could be explained with poor material quality. However, a further analysis of the screw fractures was not carried out. So far, such a high occurrence of abutment screw fracture during artificial loading was not yet reported.[19,20]

The mean fracture strength value of the zirconia implants with the all-ceramic crowns was 277 N after artificial loading (the zirconia coping and Empress 2 groups combined). However, the range of biting forces in the natural dentition varies markedly from one area of the mouth to another and from one individual to another. For the incisor region, bite forces range from 60 to 360 N.[21,22] For the premolar and molar regions, bite forces were measured in the range of 237 to 850 N.[21–24] The average maximum sustainable biting force seems to be around 800 N,[25] and the highest unilateral bite force measured was 1550 N.[22] In the context of the average biting forces presented in the literature, the long-term survival of the presented prototype zirconia implants – even in the anterior region – is questionable.

From several investigations, Körber and Ludwig[26] calculated an average maximum biting force of about 200 N as realistic. This average biting force value could be used as a reference value for detecting the minimal long-term fracture resistance (fatigue strength) for anterior reconstructions. According to some authors,[27,28] the fatigue strength amounts to approximately 50% to 60% of the initial stability (without long-term load) of a dental ceramic material because of subcritical flaw propagation. Therefore, for the safety of anterior restorations, the initial stability should be at least of a magnitude of 400 N.[29] When regarding 400 N as the mean minimal stability, the zirconia implants did not fulfill this criterion. Clinical results, however, showed that the above-mentioned minimal criteria do not respect the clinical aging of ceramic materials.[30,31] Therefore, Tinschert and colleagues[32] proposed that for clinical use, the initial stability should be at least double the fracture strength of 400 N. Neither the zirconia nor the titanium implant group fulfills this suggested requirement.

However, zirconia implant exposure to the artificial mouth had no statistically significant influence on the fracture strength of the implants. It seems that this material would not age over a period of 1.2 million cycles to such an extent that the fracture strength decreases. Whether a longer exposure to the artificial mouth might have changed the stability significantly has to be shown. In general, it is supposed that artificial loading, including thermocycling, leads to a phase transformation in zirconia from the tetragonal to the monoclinic phase called low-thermal degradation.[33,34] The increase in monoclinic phase reduces the strength, toughness, and density, possibly followed by micro–macrocracking and surface roughening.[35] Furthermore, during thermocycling, water may penetrate into the bulk material, therefore reducing the Zr-O-Zr bonds and additionally impairing the material stability.[36,37]

The fracture test values after artificial loading in our experiment were lower compared with the values of another investigation dealing with two-piece ceramic implants.[14] In the latter experiment, the fracture values for the zirconia implants restored either with Empress® 1 or Procera® Alumina (Nobel Biocare AB, Göteborg, Sweden) were 410 and 555 N, respectively. At these fracture values, the all-ceramic crowns (Empress 1 [Ivoclar Vivadent AG, Schaan, Liechtenstein], Procera Alumina) fractured, but not the implants or abutments. This is in contrast to the present investigations where, at lower fracture test values, the implants fractured. Obviously,

these differences have to be attributed to the different implant designs. In the present investigation, the implants were of cylindrical shape and the implants of the investigation by Kohal and colleagues[14] had a rootlike form, with increased material volume at the region of the implant head. This difference in material volume seems to be responsible for the different fracture values and modes.

Laboratory investigations evaluating all-ceramic systems on titanium cylinder implants[38–41] were presented in the literature. In the earlier investigations, metal components showed deformations at lower loading values (108–198 N) compared with the present investigation. Yildirim and colleagues[40], using commercially available ceramic abutments, found mean fracture load values of 280 N for alumina abutments and 738 N for zirconia abutments on non-fatigued samples. These results were corroborated by Att and colleagues,[41] who found, however, after artificial aging, a median fracture resistance of 241 and 457 N for alumina and zirconia abutments, respectively. Butz and colleagues[42] presented data on alumina and zirconia abutments, with similar fracture strength after chewing simulation for alumina abutments (239 N) but lower for zirconia abutments (294 N). The differences between the results of those investigations and our results are in the components that fractured. In our investigation, there was a fatal fracture of the implants, which would have to be removed in a clinical setting, whereas in the investigations by Yildirim and colleagues[40] and Butz and colleagues,[42] only fractures of the screws or abutments occurred. There exists the possibility of restoring the implants with new components.

CONCLUSIONS

Within the limits of this pilot investigation, the biomechanical stability of all tested prototype implant groups seems to be – compared with the possibly exerted occlusal forces – borderline for clinical use. Therefore, we do not recommend the application of the presented prototype two-piece implants in daily practice at the moment. The use of different crown materials did not influence the fracture stability of the implants.

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