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An In Vitro Comparison of Fracture Load of Zirconia Custom Abutments with Internal Connection and Different Angulations and Thicknesses: Part II

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Keywords
Ceramic abutments stability; shear stress; fracture load; implant/abutment connection, ceramics; zirconia.

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Abstract

Purpose: The purpose of part II of this in vitro study was to compare the fracture load of two-piece zirconia custom abutments with different thicknesses and angulations.
Materials and Methods: Forty zirconia custom abutments were divided into four groups as follows: group A1: 0.7 mm thickness and 0° angulations; group A2: 0.7 mm thickness and 15° angulations; group B1: 1 mm thickness and 0° angulations; group B2: 1 mm thickness and 15° angulations. As in part I, in all groups, implant replicas were mounted in self-cure acrylic jigs to support the abutments. The zirconia custom abutments were engaged in the implant replicas using a manual torque wrench. All jigs were secured and mounted in a metallic vice and subjected to shear stress till failure using a universal testing machine with a 0.5 mm/min crosshead speed with the force transferred to the lingual surface of the zirconia custom abutments 2 mm below the incisal edge. The test specimens used in this study did not include a crown. The universal testing machine was controlled via a computer software system, which also completed the stress-strain diagram and recorded the breaking fracture load. The fracture loads were recorded for comparison among the groups and subjected to statistical analysis (two-way ANOVA and Kolmogorov-Smirnov).
Results: The mean fracture load of zirconia custom abutments across the groups (A1 to B2) ranged from 432 ± 97 N to 746 ± 275 N. The angulated zirconia custom abutment exhibited the highest fracture load, which was statistically significant ($p = 0.045$). The thickness of the zirconia custom abutment also had a positive influence on the strength of the specimens ($p = 0.005$).
Conclusions: In this study, the 15° angulated zirconia custom abutments showed the highest fracture load of those investigated. The 1 mm thick zirconia custom abutments also exhibited significantly higher fracture load compared to 0.7 mm abutments.
Clinical Implications: The results of this in vitro study will help dental practitioners with their decision-making process in selecting the type of custom abutment to be used clinically.

Zirconia is a crystalline dioxide of zirconium, which offers enhanced biocompatibility.[1-3] The 3 mol% yttria-stabilized zirconia (3Y-TZP) is available for fabrication of custom abutments using computer aided design/computer-aided manufacturing (CAD/CAM),[4,5] and exhibits better mechanical properties than other zirconia combinations,[6-12] alumina oxide ceramics, and standard glass ceramics.[13-18] Zirconia abutments may promote soft tissue integration and have shown favorable esthetic outcomes compared to metal abutments.[6,19-23] However, exposure to wetness for an extended period of time, surface treatments, and grinding can have a detrimental effect on zirconia.[24-26]

Zirconia abutments with various implant/abutment connection geometries exist for different implant types. The different types of implant/abutment connections might have a critical influence on the technical outcome of zirconia abutments.[27-29]

Zirconia abutments with internal connection are available in two forms (one- and two-piece), which exhibit different resistance to loading as a result of a different distribution of the applied forces.[4] The survival rate, fracture force, and failure mode of implant abutments have been studied, and the importance of two- and one-piece zirconia custom abutments has been emphasized.[29-32] In a one-piece zirconia abutment, the abutment itself can obtain the internal connection, whereas in

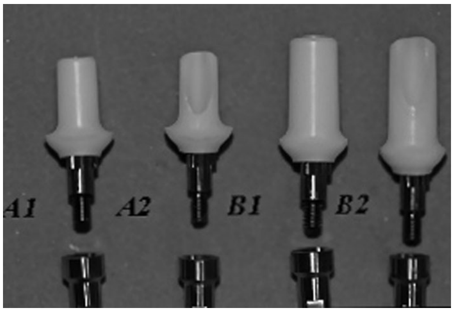

Figure 1 Procera regular platform zirconia custom abutments: (A1) straight (0° angle) zirconia custom abutment (0.7 mm thick), (A2) angulated (15° angle) zirconia custom abutment (0.7 mm thick), (B1) straight (0° angle) zirconia custom abutment (1 mm thick), (B2) angulated (15° angle) zirconia custom abutment (1 mm thick).

Table 1 Fracture load means, significant differences (SD), minimum, and maximum values (n = 10/group)

Groups	A1	A2	B1	B2
Thickness (mm)	0.7	0.7	1.0	1.0
Angulation	0°	15°	0°	15°
Mean (N)	432	587	643	746
SD	97	188	193	275
Min (N)	273	435	250	462
Max (N)	598	1022	998	1233

$p = 0.005$ between A1 and A2, B1 and B2.
$p = 0.045$ between A1 and B1, A2 and B2.

two-piece the connecting part can be either a secondary metallic component (e.g., Replace, Noble Biocare) or a secondary titanium abutment (e.g., CARES, Straumann) mounted on the implant with the abutment together by one abutment screw.[27,33]

Metallic internal connection has shown a more favorable load distribution in the connection area.[34] Significantly higher values have been achieved for CAD/CAM zirconia abutments with internal connection via a secondary titanium insert (two-piece) than for the ones with an external connection. Therefore, the use of the secondary titanium insert might have a beneficial influence on the stability of zirconia abutments.[35]

The aim of this in vitro study was to compare the fracture load of two-piece zirconia custom abutments with different thicknesses and angulations. The null hypothesis was that there is no difference between the two-piece zirconia custom abutments with different angulations and thicknesses.

Materials and methods

Forty CAD/CAM zirconia custom abutments (Procera RP [NobelReplace Select straight TiUnite RP, 4.3 × 13 mm]; Nobel Biocare, Yorba Linda, CA) were used in this in vitro study (Fig 1). The zirconia custom abutments were divided into four groups. Group A1: 0.7 mm thick, 0° angulations; group A2: 0.7 mm thick, 15° angulations; group B1: 1 mm thick, 0° angulations, and group B2: 1 mm thick, 15° angulations (Table 1).

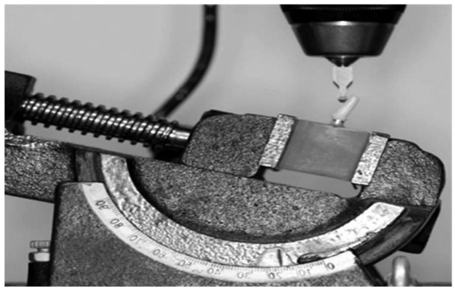

Figure 2 Metallic vice with specimen mounted and subjected to shear stress.

Specimen preparation

Forty implant replicas (10 for each group; Nob RpL RP 4.3 × 11 mm, REF 29502 LDT436479; Nobel Biocare) were placed in cubic autopolymerizing acrylic jigs (Caulk® Orthodontic Resin; Dentsply Caulk, York, PA) with dimensions of 2.5 × 2.5 × 2.5 cm^3. Each replica was attached to a laboratory surveyor (Dentsply Neytech, Yucaipa, CA) using a guide pin (Impression post, RP 4.1 mm, Nobel Biocare). The implant replicas were adjusted perpendicular to the jig's surface (90°). A water scale was used to adjust the implant replicas with the surveyor's pen.

A single operator using a surface scanner (NobelProcera™ Scanner; Nobel Biocare) scanned the implant replicas to design custom abutments digitally at the prosthodontics department (Tufts University School of Dental Medicine, Boston, MA). The surface scanner uses a laser beam to trace abutment position locator (RP 35551; Nobel Biocare), render a digitized image of the implant analog, and design the custom abutment digitally. The finish line was set and adjusted using 3D imaging software (NobelProcera® 3D GUI, Nobel Biocare). The scanned information transferred electronically to the production facility for fabricating the abutments (Nobel Biocare).

Zirconia custom abutments were then engaged to the implant replicas in the cubic acrylic jig using a manual torque wrench and torqued to 35 Ncm based on manufacturer's recommendations (Nobel Biocare). The test specimens used in this study did not include a crown. The acrylic jigs were mounted and adjusted at 30° relative to the mechanical indenter for all groups. The indenter was covered by a resilient material (Durasoft; Scheu Dental GmbH, Iserlohn, Germany). The indenter contacts the entire mesiodistal occluding surface in a contact width of approximately 2 to 4 mm. The resilient material is a co-extrusion compound material consisting of a hard polycarbonate base and soft polyester urethane, which was used to reduce localized contact stress intensities and to distribute stress over the complete testing unit, including screws and abutments.

The specimens were then mounted and secured in a metallic vice and subjected to shear stress till failure using a universal testing machine (Model 5566; Instron, Canton, MA) with a 0.5 mm/min crosshead speed with the force transferred to the lingual surface of the zirconia custom abutments 2 mm below the incisal edge (Fig 2). The universal testing machine was controlled via a computer software system (Bluehill®2 Software, Canton, MA), which also completed the stress-strain diagram and recorded the breaking loads.

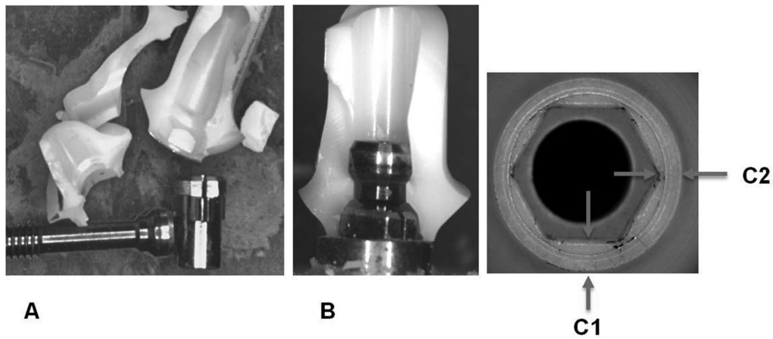

Figure 3 (A) and (B) views of the fracture pattern in the zirconia custom abutments, (C) the thickness of implant abutment connection (C1 = 0.7 mm and C2 = 0.6 mm).

Statistical analysis

Descriptive statistics were reported for each group (means, standard deviations, minimum and maximum values). A two-way ANOVA was performed to assess the statistical significance of each factor. A Kolmogorov-Smirnov test was also performed to check the normal distribution of residuals across the groups.

Results

The results of the study are shown in Table 1. Group B2 (1 mm thick, 15° angulations) fractured at a mean (SD) load of 746 (275) N, group B1 (1 mm thick, 0° angulations) fractured at a mean (SD) load of 643 (193) N, group A2 (0.7 mm thick, 15° angulations) fractured at a mean (SD) load of 587 (188) N, and group A1 (0.7 mm thick, 0° angulations) fractured at a mean (SD) load of 432 (97) N, where the numbers were rounded to the nearest 1.

The p-value of the Kolmogorov-Smirnov test was $p = 0.509$, meaning there was no evidence that the assumption of normal distribution of the residuals is violated. Based on this, a two-way ANOVA was performed and the results were as follows: The maximum fracture load was achieved in group B2. The two-piece zirconia custom abutment groups with 1 mm thickness (B1 [643 ± 193] and B2 [746 ± 275]) exhibited significantly higher fracture load compared to 0.7-mm-thick zirconia custom abutment groups (A1 [432 ± 97] and A2 [587 ± 188]), $p = 0.005$.

There were statistically significant differences between groups with different angulations ($p = 0.045$). Groups B2 and A2 with angulated abutments showed a higher fracture load than groups B1 and A1 with straight abutments, a result that was statistically significant.

Discussion

This in vitro study demonstrated that two-piece zirconia custom abutments with various thicknesses and angulations have a different fracture load under static load for standard internal connection implants. Therefore, the null hypothesis was rejected. The fracture load of all-ceramic implant abutments made from zirconia has been reported between 429 and 793 N, under load angles ranging from 30° to 60°.[4,36,37] In this study, the mean fracture load for 0° and 15° two-piece zirconia custom abutments across the groups ranged from 432 ± 97 to 746 ± 275 N, showing a strong correlation between measured fracture loads and the type of implant/abutment connection. The results of this study may not be comparable to other studies due to the different study design, testing method, variation in the angle of the applied load, the size, shape, and the material of the abutments, which all could have an effect on the final result.[4,29-32,34,36-42]

A variation of fracture pattern has been observed in alumina, zirconia, and titanium abutments with internal connection.[12,32] According to one of these investigations, implant neck distortion, fracture of the abutment, and/or fracture of both the abutment and crown were the main reasons for failure in specimens bearing titanium abutments.[12] In contrast, in this study, only the ceramic component of the abutments failed by fracture in all groups, which could have been associated with difference in force application and not including crowns in test specimens. Although in vitro studies should be as clinically relevant as possible, the absence of crowns in this study could have a weakening effect on the overall fracture load (Fig 3). Dynamic loads were used in previous studies,[12,32] whereas static loads were applied slowly with a 0.5 mm/min crosshead speed in this study, allowing higher loads before failure. This corresponds to the load in a parafunctional situation, in which higher occlusal forces than chewing are expected. In this study, mean fracture load for all zirconia custom abutments exceeded the occlusion forces reported by others.[39,40]

Artificial dynamic thermal aging was not applied to the specimens in this study due to the failure to exert a statistically significant influence on the fracture load of either straight or angulated abutments in previous studies.[41,43-46] However, it could have resulted in a lesser mean maximum applied force before failure. Nevertheless, naturally occurring forces in patients remain far below the forces recorded in these in vitro studies.[5] This study showed failure by fracture in all zirconia custom abutments with different thicknesses. However, Glauser et al reported no fracture of zirconia abutments after 4 years of clinical service.[18] In clinical situations, therefore, a plastic deformation of the metallic components is unlikely to occur; however, it is important to consider the forces that can be expected in actual clinical situations.

In this study, implant replicas were embedded in autopolymerizing acrylic resin, which is consistent with several in vitro studies.[29,41,47] However, it may be beneficial to use a material that has a modulus of elasticity and a shape and volume

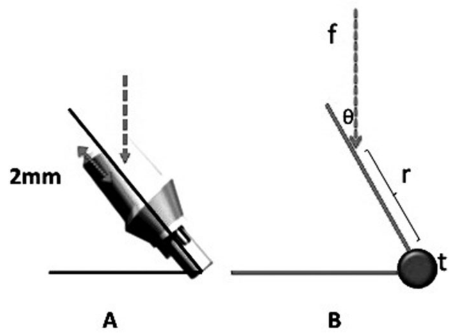

Figure 4 (A) and (B) Direction of load applied to the specimens using Instron machine. t, torque vector (implant/abutment connection); r, length of the lever arm vector (the distance between the force being applied on the abutment's surface and the implant abutment connection); f, force vector (the force applied by the Instron machine); θ, angle between the force vector and the lever arm vector.

closer to human alveolar bone, as this may have a better stress distribution effect. In this study, the torque moment played an important role on the fracture load of the zirconia custom abutments. The strength of the specimens was affected by the force applied to the specimens, the length of the lever arm connecting the axis to the point of force application, and the angle between the force vector and the lever arm (Fig 4).[48]

In addition, the thickness of the zirconia custom abutments had a statistically significant and positive influence on the strength of the two-piece zirconia custom abutments, which is in disagreement with part I of this study.[49] That could be due to the different design and fabrication (one-piece vs. two-piece) of the zirconia custom abutments. In this study, the angulated zirconia custom abutments also exhibited a higher mean fracture load compared to straight abutments, which is in agreement with Nothdurft et al's[41] findings and in disagreement with part I of this study.[49] It is possible that in two-piece zirconia custom abutments, loading forces are higher in the area of the implant/abutment connection, which is colocalized with the thinnest portion of the abutment. Current findings support the results of previous studies regarding the clinical performance of the angulated abutments.[43-45]

Further studies may be required to test different angulations and thicknesses using other implant systems. It would also be beneficial to test the specimens with artificial crowns cemented to the abutments using different types of cements. Similar in vitro studies do not replace clinical studies; therefore, their outcomes should be interpreted with caution.

Conclusions

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

1. Angulated two-piece zirconia custom abutments had the highest fracture load.
2. The thickness of the zirconia custom abutments also had a positive influence on the fracture load.

Clinical significance of the study

The results of this in vitro study will help dental practitioners with their decision-making process in selecting the type of zirconia custom abutment to be used clinically.

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