Implant Angulation Effect on the Fracture Resistance of Monolithic Zirconia Custom Abutments: An In Vitro Study

Running Title

Fracture Resistance of Angulated Ceramic Abutments

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Abstract

Purpose: To investigate the fracture resistance and performance of zirconia when employed for the fabrication of implant abutments with different angulations, simulating anterior maxillary oral rehabilitation.

Materials and Methods: Forty-five monolithic zirconia custom abutments of internal conical implant connection were CAD/CAM designed and fabricated. The specimens were divided into three groups (n=15/group) according to implant- to- abutment angulation. The angulations used were; 0°, 15°, and 25°.

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The abutments were loaded until failure at 135° using the Universal Testing Machine (Instron, Canton, MA). Collected data were statistically analyzed using one-way ANOVA and post-hoc Tukey test.

Results: Mean (±standard deviation) load at fracture of the zirconia abutments for the three groups were 962.37 ±93.81 N (Gr15) > 718.25 ±93.71 N (Gr25) > 534.05 ±133.77 N (Gr0). Statistically significant difference ($p<0.0001$) was found between all groups; Gr0 vs. Gr15, Gr0 vs. Gr25, Gr15 vs. Gr25.

Conclusions: Contrary to expectations, the non-angulated monolithic zirconia abutments presented the lowest fracture resistance values. Angulating the abutments 15 or 25 degrees, following the palatal resorption pattern of the premaxilla, significantly increased the in vitro fracture resistance.

Keywords:

monolithic; zirconia abutment; implant; fracture resistance; angulation; bioengineering mechanics

Single implant rehabilitation in the esthetic zone presents a great clinical challenge in order for predictable functional and esthetic long-term success, while the increased clinical incidence of abutment screw fractures at the site of the maxillary central incisor calls for further investigation of its limitations and optimization of the treatment design.[1-3] The high esthetic demands of the anterior zone have favored the selection of ceramic materials over the original use of titanium, as a grayish/dark optical effect has been observed at the periimplant soft tissue around the titanium implant/abutment interface, with a greater incidence for thin periodontal biotype.[4-6]

Despite the enhanced material properties of yttria-stabilized zirconia ceramic, such as the high flexural strength, fracture toughness and the unique transformation-toughening mechanism that resists crack propagation[7,8], failures of zirconia abutments have been observed clinically.[9,10] Additionally, lower fracture resistance has been reported in vitro when combining lithium disilicate or zirconia crowns with zirconia abutments, compared to cementing them on titanium

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abutments.[11] Hydrothermal degradation and cyclic fatigue in aqueous environment, as seen during oral function, magnifies mechanical failures of the material, with progression of spontaneous transformation of the tetragonal phase into monoclinic phase.[7,12] Various clinical and mechanical parameters have been investigated in order to minimize the fracture incidence under function, with recommendations on the thickness and design of the zirconia abutments, as well as the fabrication process and treatment of the material.[13,14] The combination of implant components from different systems may introduce a risk factor for the performance of zirconia abutments as well, as indicated by the lower fracture resistance of monolithic zirconia abutments when combined with a nonproprietary implant system.[15] Regarding zirconia thickness, a minimum of 0.7 mm axial wall thickness has been suggested for zirconia abutments to withstand occlusal forces[16], while 0.5 mm- thickness copings have been suggested to withstand warpage[17] and fracture even on posterior sites.[18]

The internal implant/abutment connection has been favored over the original external hexagon for its more favorable stress distribution of the applied functional stresses. The stresses are applied not only to the mechanical components of the system and, thus, affecting the mechanical survival of the system, but they are distributed to the surrounding bone as well.[19,20] The latter holds great significance for the biologic and esthetic long-term success, which relies intimately on the maintenance of the crestal bone level and on the interrelated soft tissue profile.[4,21,22]

Clinically, proper three-dimensional implant position guided by the desired restoration position has been emphasized for controlled esthetic outcomes.[1] Anatomic limitations are often encountered in the premaxilla region that may lead to a palatally angulated position of the implant apex. This is attributed to the physiologic bone remodeling with a palatal resorption pattern in the premaxilla, and it can be further influenced by the frequently encountered thin

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buccal plate of less than 1 mm.[23-25] Prosthetic customization of the implant abutment can accommodate for the osseous topography in relation to the prosthetic crown, with a high clinically observed 2-year survival of titanium abutments often angulated in the range of 5-30º.[26]

Only a few in vitro studies have reported on the influence of the implant/abutment angulation on the fracture resistance of internal connection zirconia abutments.[16,27-29] One study found that 15° angulation of one-piece zirconia abutments significantly decreased the fracture strength, while increasing the thickness from 0.7 to 1 mm did not alter fracture strength significantly.[16] Similarly, another study found reduction of fracture strength for 20° angulation compared to straight (0°) one-piece zirconia abutments, but the difference was not statistically significant.[28] On the contrary, it was earlier reported that one-piece zirconia abutments with 20° angulation presented a higher fracture resistance compared to 0° angulation (straight abutments)[27], although in vitro results on angulated two-piece internal connection zirconia abutments are not in agreement with the aforementioned observations.[29]

The purpose of the present in vitro study was to further investigate the relation between the implant-abutment angulation and the fracture resistance of zirconia ceramic abutments, as no conclusive results exist in the literature on the angulation threshold of internal connection one-piece zirconia abutments for the successful use in the single implant rehabilitation of the anterior maxilla. Three implant-abutment angulations were evaluated in order to identify the threshold of the implant-abutment angulation for the specific implant system tested. The null hypothesis was that the increase of the implant-abutment angulation would not affect the fracture resistance of the custom zirconia abutments.

Material and methods

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The study was designed simulating implant rehabilitation of a maxillary central incisor with different transverse positional relation of the alveolar bone and the prosthetic restoration, following the palatal oriented pattern of alveolar bone resorption following extraction of the central incisor.[30]

Three clinical scenarios were selected; implant placement that follows the original root configuration (0º), and implant placement that deviates palatally from the original root configuration either moderately (15º) or more severely (25º). The immediate aim was to investigate if this deviation will result to an implant/ prosthetic complex where monolithic zirconia abutments can withstand the occlusal load. The secondary aim is to start developing a prosthetic deviation protocol based on survival and success of the prosthetic rehabilitation.

Forty-five internal conical connection (4.3 mm; NobelActive RP; NobelBiocare USA, LLC, Yorba Linda, CA) zirconia abutments were CAD/CAM fabricated and divided into three groups (n=15) (Fig. 1). Group0 (Gr0) simulated an implant positioning that allows for the fabrication of straight abutment (deviation of the long axis of the abutment from the long axis of the implant of 0º). Group15 (Gr15) simulated a moderate implant angulation (deviation of 15º), and Group25 (Gr25) simulated more severely compromised position with 25º angulation.

A gypsum model that incorporated the implant replica was scanned using the NobelProcera Scanner (NobelBiocare USA, LLC, Yorba Linda, CA). One-piece zirconia abutment was digitally designed with implant-abutment angulation of 0º (Gr0) using the NobelProcera software (NobelBiocare USA, LLC, Yorba Linda, CA). The digital file was triplicated, and the Gr0 design was digitally modified to 15º (Gr15) and 25º (Gr25) using the NobelProcera software's tool that allows for axial modification of the long axis of the abutment, simulating the clinical-based procedures of abutment design based on the implant positioning and the desired position of the prosthetic restoration.

A wax platform was added on the lingual surface at a distance of 2 mm cervically from the incisal edge providing a mating flat surface for the 1 mm diameter, flat-end conical tip of the mechanical indenter, in order to ensure proper loading calculations by the testing machine

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software (Bluehill 2 Software, Canton, MA) based on the formula Stress=Load/Area for tension or compression along the longitudinal beam axis. The angle created between the custom abutments and the universal testing machine indenter was set to 135º in order to simulate the Angle Class I anterior dental occlusal relationship.[13,28,31] The abutments were scanned and fifteen identical zirconia abutments were milled from the company for each of the three groups. No external surface or further treatment was performed to the milled zirconia abutments.

Forty-five zirconia custom abutments were secured on implant replicas, which were further embedded in auto-polymerizing acrylic resin (Samplkwick Acrylic System, Buehler) according to the ISO Norm 14801:2016(E) that requires 3 mm implant neck exposure in order for an increased torque effect.[32] A dental surveyor (Ney Surveyor, Dentsply International) was used to embed each implant replica into auto-polymerizing acrylic resin. The arm of the surveyor was replaced with the stainless steel mechanical indenter of the Instron Universal Testing Machine (Model 5566; Instron, Canton, MA). Each abutment was securely attached to the indenter using a custom-made acrylic resin transfer jig (Fig. 2A, B).

A manual torque wrench (NobelBiocare USA, LLC, Yorba Linda, CA) was used to torque the abutment screws to the implant replicas at the recommended torque of 35 N, and they were re-torqued after 10 minutes.[33] The abutments were loaded until failure as determined by audible crack and/or visual observation, with a 0.5 mm/min crosshead speed transferred through the indenter at a 135º angulation to the long axis of the abutment. The maximum load and load-at-failure were recorded and generated by the software (Bluehill 2 Software, Canton, MA).

A priori power analysis and sample size and calculations were performed based on existing data of similar in-vitro studies (G*Power software).[13,28,34,35] Statistical analysis of the data was performed using XLSTAT (Addinsoft, New York, NY) software. One-way analysis of variance

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(ANOVA) with the level of significance set at α = 0.05 was used to analyze the data and Tukey's post-hoc pairwise multiple comparisons were performed in order to determine if statistically significant differences existed among the three implant-abutment angulation designs of the abutments.

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Results

A summary of the statistically analyzed results of the load to fracture and maximum load in vitro testing of the three experimental groups is shown in Table 1. Gr15 presented the highest fracture load, compared to Gr0 and Gr25. The load to fracture values of Gr15 ranged between 822.68 N to 1168.79 N. The values of Gr0 ranged between 233.68 N to 852.75 N, while the ones for Gr25 ranged between 589.87 N to 875.19 N. The mean (±standard deviation) loads at fracture of the custom zirconia implant abutments for the three experimental groups were 962.37 ±93.81 N (Gr15) > 718.25 ±93.71 N (Gr25) > 534.05 ±133.77 N (Gr0). The experimental groups presented the same statistical significance relationships regarding the maximum load values recorded. The mean (±standard deviation) maximum loads of the custom zirconia implant abutments for the three experimental groups were 1167.90 ±130.64 N (Gr15) > 957.07 ±114.47 N (Gr25) > 762.70 ±109.60 N (Gr0) (Table 1).

One-way ANOVA analysis indicated that there was statistically significant interaction between the implant-abutment angulation and the fracture load ($p < 0.0001$, $F=58.55$) of the zirconia abutments when loaded to fracture. The pairwise comparison differences of any group

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combination were found statistically significant for both load -to -fracture and maximum load data ($p<0.0001$); Gr0 vs. Gr15, Gr0 vs. Gr25, Gr15 vs. Gr25.

All the abutments remained attached to their corresponding implant replicas after the performance of the load-to-failure testing, without abutment mobility observed. The use of the corresponding screwdriver was needed in order to detach the abutments. The failure was observed as fracture at the internal conical joint for all the specimens, while the fractured particles were remaining inside the implant replica. In addition, circular abrasions marks were observed on the internal connection of the abutments (Fig. 4).

Discussion

The present in vitro study reports on the influence of different implant angulations in relation to the prosthetic-driven abutment orientation on the fracture resistance of monolithic zirconia abutments. The null hypothesis is rejected as the abutment angulation is found to be a statistically significant variable on the fracture resistance. The abutments with 15º long axis deviation from the implant axis presented the highest fracture resistance between all groups, while the abutments with 25º deviation presented significantly higher fracture resistance compared to 0º deviation. The differences were statistically significant between all group combinations (Table 1).

Implant placement in the premaxilla may deviate from the original dental root configuration. Increased incidence of thin buccal plate in extraction sockets and physiologic buccal remodeling found at delayed implant placement protocols are contributing to the need for either osseous augmentation procedures or deliberate deviation of the implant body and apex in order to achieve the optimum prosthetic orientation.[22-25] As a result, angulating the prosthetic

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abutment compensates for the implant long axis deviation. Clinically, abutments have been treatment planned more often with an angulation between 5º and 30º, and a high survival of angulated titanium abutments has been reported.[26] The performance and reliability of ceramic angulated abutments remains a crucial clinical question.

In regard to the results of this study, differences of the abutment design and thickness between the three groups, especially on the cervical part of the zirconia abutments, may have contributed to the results. The resulting abutment designs present apparent and inevitable differences in both form and thickness. As seen in green (Fig. 5), every zirconia custom abutment of the specific CAD/CAM system is fabricated with the same internal connection joint, with an extension of a certain form and size into the cervical area of the body of the abutment. The variability lies on the additional material surrounding this standardized form, which is directly influenced by the customization of the abutment design such as the angulation of the long axis. Further investigation of the influence of the form and thickness of these areas, besides the long axis inclination, would further shed light on the proper zirconia abutment designs in order to control the long-term success. Furthermore, our results present inner-group data variation that is reflected by the 93.71 to 133.77 standard deviation range. As observed by the scatter graphs (Fig. 3. A, B), it is of interest that the data for Gr0 are much closer to the mean value than the data for Gr15 and Gr25, indicating increased reliability of the results for Gr0. In general, the standard deviation for all the groups may be attributed to intrinsic characteristics of the material, to manufacturing factors during milling of the abutments, and to operator error during the use of the universal testing machine as no specialized personnel was conducting the testing.

In accordance with previous observations, abrasion marks were observed at the internal connection of the zirconia abutments[9,36], and the failure was located on the internal conical connection below the implant shoulder.[13,15,27,37] In a previous study, the fracture was reported at

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the abutment level even when a zirconia crown was cemented on the abutment.[11] The conical connection was always fractured close to its terminal area and always below the implant platform level, while the fracture was in form of both irregular particulates but also of ring or semi-lunar formations (Fig. 4). Interestingly, the conical connection near the level of the implant platform, as set by the specific system, is close to 0.6 mm (Fig. 5). The fracture was always observed below that level, which corresponds to an area of even less thickness of material. Similar measurements at the internal connection have been previously reported for another system[16], while these findings are of high importance as new zirconia abutment designs of increased thicknesses at the internal joint could attribute to improved performance.

Additional studies are needed for the understanding of the failure behavior of the material under loading, but it is of particular clinical interest that although a significant part of the internal connection was fractured there was no detachment or apparent looseness of the abutment. Similarly, absence of abutment mobility has been previously observed after thermal cycling and load-to-failure of one-piece zirconia and titanium abutments, in contrast to zirconia external and two-piece internal connection abutments.[37] Those observations are of great clinical importance, as one-piece internal connection zirconia abutments may have failed without presenting any apparent indications and, thus, the patient and the clinician may not become aware until further progress of the fracture.

A limitation of the present study is that fatigue from thermal cycling was not evaluated. Although thermal cycling of a clinically relevant regime (5-55^{o} C) has been found to increase the amount of tetragonal-to-monoclinic phase transformation as investigated on the surface of the zirconia specimens, there have been speculations that the transformation zone may be limited superficially and, thus, is not affecting the strength of the material.[38] Nevertheless, future

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evaluation under combined thermal cycling and mechanical loading over 1,000,000 cycles could provide additional information on the long-term performance.[20]

The lack of uniform methodology may play a significant role on the observations of in vitro fracture studies. In regards to the orientation of zirconia abutments on the universal testing machine, various loading angulations have been used such as 30º [15,16,29], and 135º /45º.[13,28] The 30º angular loading can be justified by the ISO 14801:2016.[32] However, the aforementioned International Standard is advocated for "comparing implants with different designs and sizes". It states that it is not intended for testing "the fundamental fatigue properties of the materials from which the endosseous implants and prosthetic components are made" and cannot predict the in vivo performance. In addition, ISO 14801 refers to function loading, which is more applicable to cyclic loading than load-to-fracture.[32]

The establishment of a uniform protocol is of high importance in order for proper inter-study comparisons. This study attempted to report on the initial failure observation on a macro-level rather than of catastrophic failure, as crack formation and material deformation that leads to audible crack perception denotes a clinically relevant prosthetic failure, which jeopardizes the long-term success of the prosthesis. However, reporting this initial moment of clinically relevant macro-failure can incorporate observation errors due to the increased subjectivity of interpretation and reporting between different operators. Acoustic monitoring for determination of breakage or failure of dental ceramics, rather than depending on the sharp drop in load, was originally advocated as more reliable method based on in-vitro fracture testing experience of cemented ceramic crowns before the use of the contemporary zirconia ceramic material.[39] In order to minimize this potential limitation, the objective maximum load values that were automatically registered by the software were also reported and analyzed in the present study.[34,40] For the present study, both the load-to-failure and maximum load values produced the

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exact same statistically significant results, which assure the relationship between the parameters tested.

Zirconia abutments with all three angulations tested can successfully withstand the occlusal load transmitted to the anterior maxillary region. However, clinical observations on zirconia abutment fracture incidence cannot be ignored and various design and treatment parameters have been investigated in order to establish protocols that enhance longevity with long-term success.

Conclusions

The angulation between the long axis of the implant and the zirconia abutment significantly affected the resistance to fracture of one-piece zirconia abutments with internal connection. In addition, this study showed that the implant-to-abutment angulation of 15º presented statistically significant higher fracture resistance values than 0º and 25º. Deviating the implant apex 15º or 25º palatally present an acceptable treatment in regards to the fracture resistance of monolithic zirconia abutments.

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Table 1 Means and Standard Deviation of (A) Load at Fracture values (N), and (B) Maximum Load values (N)

(A)

Implant-Abutment Angulation	Load At Fracture Means [N]	SD	Lower bound (95%)	Upper bound (95%)
Gr15	962.365[a]	93.811	905.699	1019.030
Gr25	718.246[b]	93.713	661.581	774.911
Gr0	534.045[c]	133.775	477.380	590.711
Pr > F	< 0.0001			
Significant	Yes			

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(B)

Implant-Abutment Angulation	Maximum Load Means [N]	SD	Lower bound (95%)	Upper bound (95%)
Gr15	1167.899[a]	130.635	1106.112	1229.685
Gr25	957.074[b]	114.474	895.287	1018.861
Gr0	762.696[c]	109.599	700.909	824.483
Pr > F	< 0.0001			
Significant	Yes			

* Values with different superscripted letter (a, b, c) denote statistical significant difference at $\alpha = 0.5$ between groups.

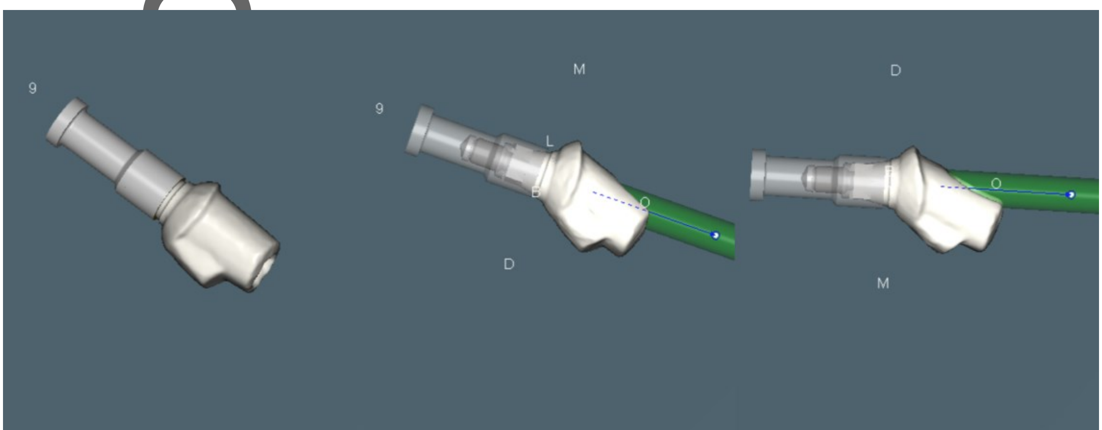

Fig. 1. Digital experimental designs of Group0, Group15 and Group25, as shown from left to right.

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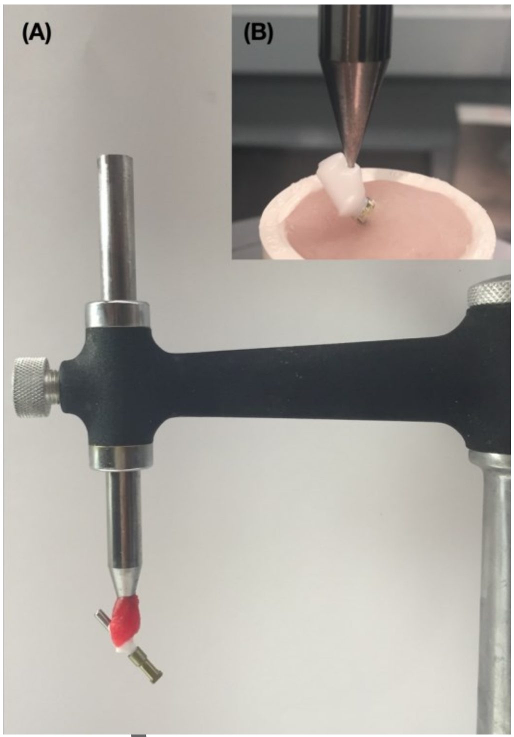

Fig. 2. (A) Custom-made mounting transfer jig, and (B) the specimen positional relation to indenter.

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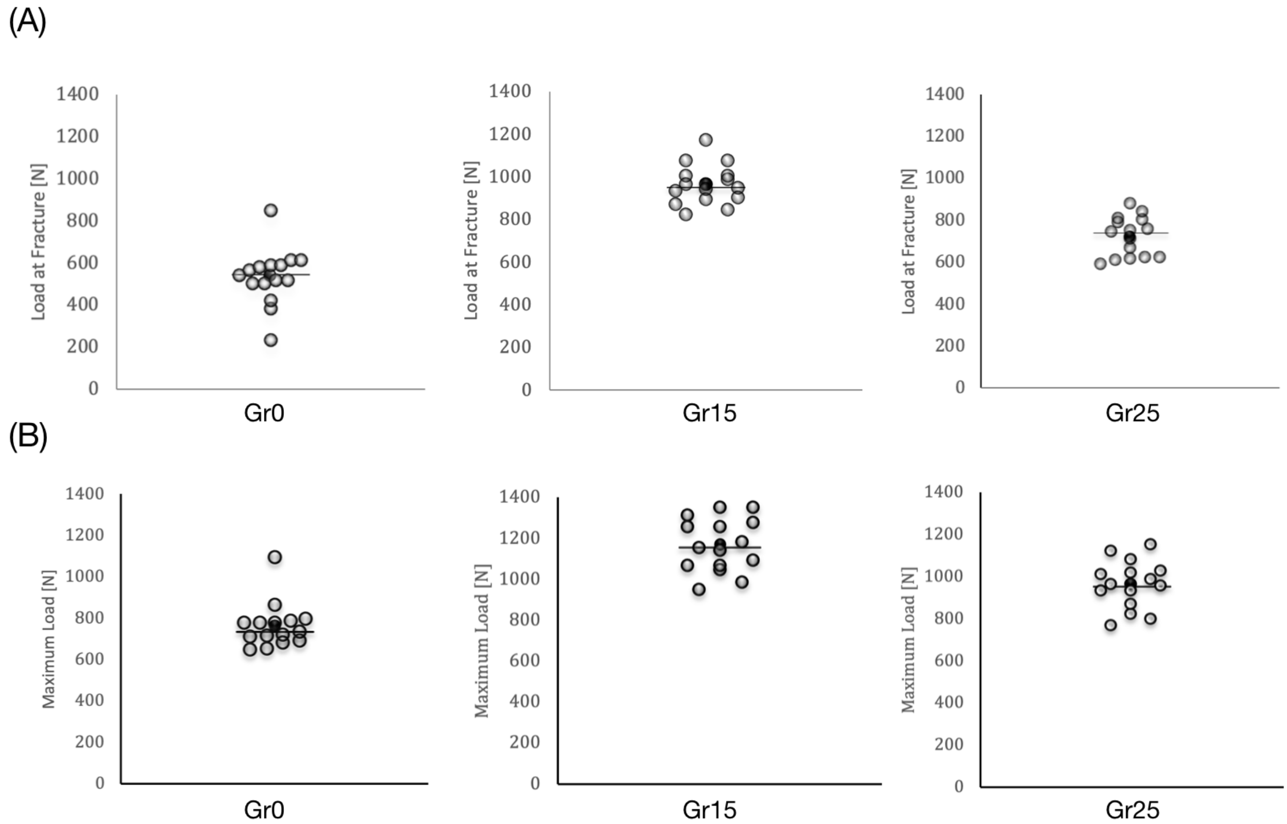

Fig. 3. Scatter graphs of (A) Load at Fracture (N) (Mean; solid black dot, Observations; circles, Median; black line) and (B) Maximum Load (N) (Mean; solid black dot, Observations; circles, Median; black line) data distribution of the observations and means of the experimental groups

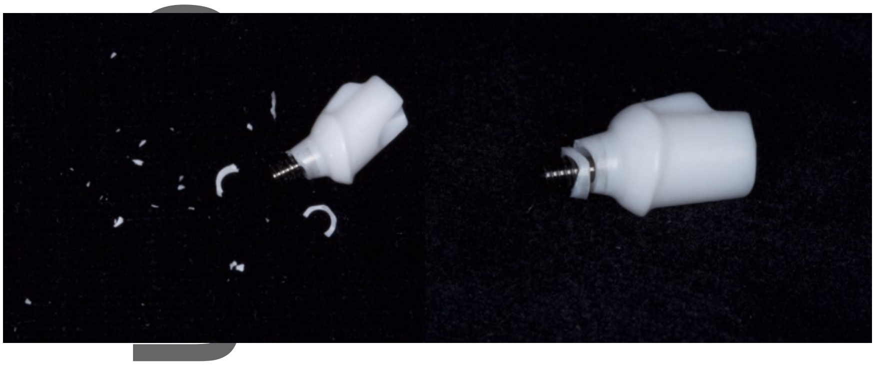

Fig. 4. Representative fracture pattern of specimens.

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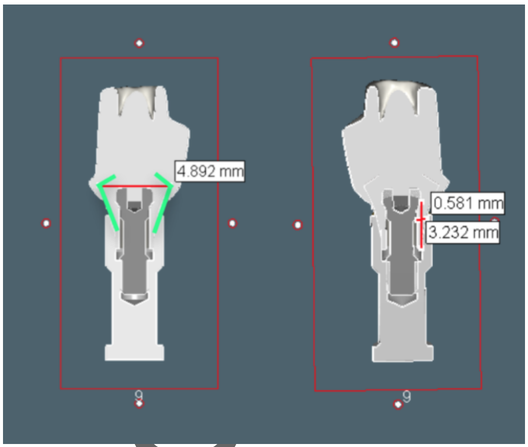

Fig. 5. Digital measurements of the internal joint connection and standardized cervical dimensions using Procera software (NobelBiocare).

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