Mechanical Stability of Zirconia Implant Abutments Supporting Cantilevered Fixed Dental Prostheses After Fatigue Loading

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Abstract

Purpose: To evaluate the mechanical stability and complication rates of titanium (Ti) or zirconia (Zr) abutments restored with cantilevered fixed dental prostheses (cFDPs) when supported by 1 or 2
implants. **Materials and Methods:** A total of 32 samples were fabricated. Half of the samples received 1 implant, and the other half received 2 implants (Bone Level Implant, $\phi$ 4.1, 13 mm, Straumann) to simulate the clinical situation of 2 or 3 missing maxillary incisors, respectively. Each group was divided into two subgroups ($n = 8$). Ti-I and Ti-II groups received Ti abutments (Anatomic Abutment, Straumann) supporting 2- or 3-unit metal cFDPs, respectively, while Zr-I and Zr-II groups received Zr abutments (IPS e.max Anatomic Abutment, Straumann). Following the cementation of cFDPs using resin cement (Multilink Automix, Ivoclar Vivadent), the samples were subjected to thermomechanical fatigue load and were subsequently loaded until fracture in a universal testing machine. Following the static loading test, stereomicroscopic analyses (Carl Zeiss) were done to identify the weakest component of the cFDP, abutment, and implant assembly. Mann-Whitney $U$ test was used to evaluate the effect of the number of supporting implants and abutment material on fracture strength values, and the level of statistical significance was set at 5% ($\alpha = .05$). **Results:** All specimens survived the aging, and no screw loosening or fracture was recorded. The mean fracture strength values were 226 N ($\pm 26.45$), 551.12 N ($\pm 82.19$), 601 N ($\pm 41.51$), and 664.5 N ($\pm 37.59$) for Zr-I, Zr-II, Ti-I, and Ti-II, respectively. The difference between fracture strength values of Ti and Zr groups was significant in favor of Ti abutments ($P < .001$). The number of supporting implants showed a significantly positive effect on the fracture strength of Zr abutments. **Conclusion:** Zirconia abutments demonstrated lower fracture strength values than titanium abutments independent from the number of supporting implants when used under cFDPs. Two-implant supported cFDPs with zirconia abutments have the potential to withstand physiologic forces applied in the anterior region. *Int J Prosthodont* 2021. doi: 10.11607/ijp.6700

**Introduction**

Clinical evidence indicates that the replacement of single missing teeth with dental implants in the esthetic zone can be successful in both esthetic and functional aspects (1, 2). On the other hand, reconstruction of an extended edentulous area in the maxillary anterior zone by placing implants is not considered as a predictable procedure regarding the esthetic outcomes (3, 4). Related to the decrease in both vertical and horizontal dimensions of bone and soft-tissue volumes following multiple tooth loss, alveolar ridge gets flattened and this may lead to difficulties in placing adjacent implants.
with adequate inter-implant distance without complex soft and hard tissue managements. (2) To avoid
the possible esthetic complications by placing adjacent implants in the esthetic zone, the reduction of
the number of implants and to restore the edentulous space with implant-supported cantilevered FDPs
(cFDP) may be considered.

Implant-supported cFDPs allow a simpler and less costly rehabilitation (5). The clinical
outcome of cantilevered implant-supported FDPs has been evaluated in a number of studies. While
one study reported higher technical and biological complication rates (6), other studies showed similar
outcomes to those of non-cantilevered implant-supported FDPs (7-9). In a systematic review by
Romeo et al (2012), cantilevered implant-supported FDPs was considered as a reliable treatment
option that does not increase the complication rate (10). Despite the promising current evidence,
unfortunately, there is still a lack of information comparing the effect of number of supporting implants
on mechanical behavior of zirconia abutments under implant-supported cFDPs in the anterior region.

In the maxillary anterior region, success depends not only on successful osseointegration but
also on the harmonious integration of the restoration with neighboring teeth regarding both pink and
white esthetics aspects. Implant abutments are usually being fabricated from titanium because of its
well-documented biocompatibility and mechanical properties (11). One major disadvantage of metal
abutments is their gray gingival discoloration that was reported in several studies (12, 13).

Ceramic abutments were developed as an alternative to metal abutments to overcome
esthetic problems originating from grey gingival discoloration and they offer several clinical
advantages, including well-documented esthetic benefits, less bacterial adhesion and plaque
accumulation compared to titanium abutments (14), as well as biocompatibility similar to titanium
abutments (15, 16). Even though the 5-year survival rate of implant abutments under fixed
reconstructions was reported to be similar for ceramic and metal abutments (16), ceramic abutments
have a major shortcoming which is their brittleness and less resistance to tensile forces. Implant
supported reconstructions with cantilever extensions reportedly create higher non-axial forces on the
implant and implant components near the cantilever extension (17). Accordingly, an in vitro study with
a clinically relevant design would provide essential information about the use of zirconia abutments
under implant-supported cFDPs. Hence, the aim of this in-vitro study was to compare the fracture
strength of zirconia (Zr) and titanium (Ti) abutments supporting cFDPs using a device simulating oral
environment and cyclic loading. Further, to evaluate the effect of number of supporting implants on the failure of both reconstructions and implant components and to obtain data about the weakest component of the entire system following maximum fracture resistance test was aimed. The tested hypotheses were: a) Fracture strength values of implant abutments under cFDPs would be influenced by the number of supporting implants and; b) Titanium implant abutments would have higher fracture strength values compared to zirconia abutments when supporting the cFDPs.

Materials and Methods

Thirty-two samples, simulating two clinical situations as 2 or 3 missing maxillary incisors, were fabricated. Half of the samples received one implant in the position of the right central incisor to support 2-unit FDP frameworks with a cantilever extension to the lateral incisor area (Figure 1A, B), and the other half received 2 implants in two central incisor positions to support 3-unit FDP frameworks with a cantilever extension to the lateral incisor site (Bone Level Implant, ∅ 4.1, 13 mm) (Institut Straumann AG, Basel, Switzerland). (Figure 1C, D). Each group was divided into 2 subgroups of 8 specimens each (Ti-I, Ti-II, Zr-I, Zr-II) (n=8). Groups Ti-I and Ti-II received one-piece internal connection titanium abutments (Anatomic Abutments, Institut Straumann AG, Basel, Switzerland) supporting cantilevered 2- and 3-unit FDPs, respectively, whereas Zr-I and Zr-II groups received one-piece internal connection zirconia abutments (IPS Emax Anatomic Abutments, Institut Straumann AG, Basel, Switzerland).

A maxillary Typodont (Frasaco GmbH, Tettnang, Germany) model was used to fabricate the samples in order to obtain a clinically relevant master model with standard inter-implant space and favorable angulation of the implants. The Typodont model was modified by removing central incisors and right lateral incisor. The lateral incisor site was re-shaped with wax to create an edentulous crest form. A pattern resin (Pattern Resin, GC International AG) duplication model was fabricated to be used as a master model. Implants were placed using a surveyor in the center of alveolar sockets and fixed with the same pattern resin. The titanium and zirconia abutments were placed on the implants and abutment level digital impressions were taken using a laboratory scanner (Lava, 3M ESPE, MN, USA) for every study group. The 2- and 3-unit cFDPs were designed based on the digital impressions with a 6-mm cantilever length, a 0.5-mm restoration wall thickness and a 9-mm² connector area. One-implant supported (n=16) and 2-implant supported (n=16) Cr-Co cFDP frameworks (Wirobond C+, Bego,
Bremen, Germany) were fabricated by using a laser sintering technique (EOS M290, EOS, Krailing, Germany). The abutments were removed and transfer copings were placed on the master model. Impressions with a custom-made trays and polyether impression material (Impregum Polyether, 3M ESPE, St. Paul, MN, USA) were taken and used as an index while fabricating thirty-two standardized samples. Implant replicas were embedded in polyurethane molds suitable for chewing simulator with auto polymerizing acrylic resin (Technovit 4000, Heraeus Kulzer, Wehrheim, Germany) at an angle of 30° to the horizontal plane to simulate clinical conditions (Figure 2)(18-22). The resin that was used had a modulus of elasticity of approximately 12 GPa which approximates that of human bone (18 GPa).

The zirconia and titanium abutments were fixed on the implant replicas using titanium screws and torqued to 35 Ncm according to the manufacturer’s recommendations. After 1 minute, the occlusal screws were retightened. The inner surfaces of the frameworks were sandblasted uniformly with 50 μm aluminum oxide and cleaned in an ultrasonic unit for 1 minute. The Zr and Ti abutment outer surfaces were sandblasted with 30 μm aluminum oxide under 2 bars of pressure for 10 seconds. The abutments were then rinsed thoroughly and dried with oil-free air. A universal primer (Monobond Plus, Ivoclar Vivadent AG, Liechtenstein) were applied to both abutment and restoration inner surfaces. Then, all crowns were definitively cemented using a resin luting cement (Multilink Automix, Ivoclar Vivadent AG, Liechtenstein).

The detailed protocol for thermo-mechanical fatigue test was published elsewhere (23). Briefly, all of the samples were exposed to 1,200,000 cycles of thermo-mechanical fatigue in a computer-controlled dual-axis chewing simulator (Willytech, Munich, Germany) to simulate the clinical function. The force was applied 3 mm below the incisal edge on the palatal aspect of the cantilever area at a frequency of 1.6 Hz using a ceramic ball with a 6 mm diameter (Steatite Hoechst Ceram Tec, Wunsiedel, Germany). Given to its spherical shape, the contact of the ceramic ball to the restoration surface was restricted to one point. It was assured that there is not interference with the ceramic ball from other parts of the restoration surfaces (i.e. connector, neighboring tooth surface). The ceramic ball has a Vickers hardness that is similar to that of enamel. A force of 49 N was chosen to simulate a load within clinical range. During testing, all specimens were subjected to simultaneous thermal cycling between 5°C and 55°C for 60 seconds each, with an intermediate pause of 12 seconds,
maintained by thermostatically controlled liquid circulator (Haake, Karlsruhe, Germany).

The samples were examined under digital microscopy (Carl Zeiss, Aalen, Germany) to identify any possible complications following the chewing simulation such as cracks on both abutments and frameworks, screw loosening and de-cementation and finally, all survived specimens were loaded compressively in a universal testing machine (Zwick/Roell Z010, Zwick, Ulm, Germany) with force application positioned in the same position with fatigue loading at an angle of 30° to the implant axis and cross-head speed of 1mm/min (Figure 3A, B). A 0.5-mm-thick thin foil (Dentaurum, Ispringen, Germany) was placed to ensure homogenous stress distribution. The applied force was graphically recorded on an x-t recorder (Zwick testXpert V.7.1, Zwick). The specimens were loaded until failure with static load, and the failure load was registered as soon as fracture load decreased by 20% of the maximum load (Fmax). Following the static loading, each specimen was examined with a stereomicroscope (Carl Zeiss, Aalen, Germany) once again to locate and determine the mode of failure. Firstly, the samples were examined to determine whether there is a plastic deformation or fracture in any component of the implant-abutment-framework-acrylic resin assembly. The fractured samples were further analyzed to determine the localization of the fracture. In the samples with abutment fracture, the levels of the fracture were divided into fracture above or below the implant shoulder. Furthermore, for the two implant-supported samples, the fracture localization differentiated as first and second abutment according to the proximity to the cantilever area.

The fracture strength values were analyzed using SPSS Version 20.0 (IBM SPSS Statistics, Chicago, IL, USA) and R programming language (24). Normality assessment of numerical variables were evaluated using graphical approach and Shapiro-Wilk's normality test. Mean fracture resistance values were compared between number of supporting implants (one vs. two) for each material subgroups. Similar comparison was conducted between materials (Ti vs. Zr) for each number of supporting implant subgroups. Although mean fracture resistance values were found to be normally distributed for each subgroup, nonparametric tests were preferred due to very small sample sizes (i.e. 8 samples in each group) The rationale behind this selection was normality tests and graphical evaluations might poorly perform on small samples and such samples most often pass normality tests due to little power of detecting non-normally distributed data (25). Between group comparisons were
evaluated using Mann-Whitney U test due to small sample sizes, and the level of statistical significance was set at 5% ($\alpha = 0.05$).

**Results**

All samples survived 1,200,000 cycles of thermo-mechanical fatigue loading simulating 5 years of clinical function. No abutment, implant replica or restoration showed visible cracks or deformation. No mobility of the superstructure or screw loosening was detected.

The mean fracture strength values were 601.0 ± 41.51 N (Ti-I), 664.5 ± 37.59 N (Ti-II), 226.0 ± 26.45 N (Zr-I) and 551.2 ± 82.19 N (Zr-II) as shown in Table 1. The mean fracture strength values were significantly higher for Ti abutments when compared to Zr abutments for both subgroups of number of supporting implants ($p = 0.003$ and $p < 0.001$). However, Ti abutments had approximately 2.7 times higher resistance than Zr abutments when one implant was used. Accordingly, Ti abutments with two implants had significantly higher fracture strength values than Zr abutments; however, the difference between materials was not high as in one implant supported samples. An overall comparison between each clinical scenario is graphically shown in Figure 4.

The samples from titanium abutment groups showed similar load-displacement curves for both one- and two-implant supported samples with an elastic deformation region and a plastic deformation region (Figure 5A). On the other hand, the samples from zirconia groups showed load-displacement curves with elastic deformation until the elastic limit and a sudden decrease at the failure point (Figure 5B).

The failure mode for every group is described in detail in Table 2. In groups Ti-I and Ti-II, the failure was presented by plastic deformation of implant–abutment assembly and acrylic resin fracture (Figure 6A). In all samples of Ti-II group, the plastic deformation was accompanied by the acrylic resin fracture. Whereas, except for 2 samples in Ti-I group, the failure was represented by plastic deformation of implant–abutment assembly (Figure 6A). All frameworks remained intact in all groups after the fracture strength test.

In groups Zr-I and Zr-II, the predominant reason for failure was catastrophic fracture of the abutment except for one sample in the Zr-I group, which demonstrated a perpendicular crack line above the implant shoulder, and no catastrophic failure. In all samples with catastrophic fracture, the fracture line was located above the implant shoulder. Furthermore, seven samples from the Zr-II
samples showed failure of the abutment in immediate proximity to the cantilever site (Figure 6B) while the second abutment showed no visible fracture above the implant level. In one sample cementation failure on the second abutment was observed. The fracture line was located at the abutment neck close to the screw level (Figure 6C). No screw fracture or loosening was detected in these groups (Figure 6D).

The implant abutment was identified as the weakest component for one- and two-implant supported zirconia abutment groups, whereas the implant-abutment assembly and the embedding resin were identified as the weakest component in the Ti-I and Ti-II groups.

**Discussion**

In the present study, two-implant supported cFDPs based on zirconia abutments exhibited significantly higher fracture strength values than the one-implant supported ones. In titanium abutment groups a similar trend was observed, yet, the difference was negligible. Thus, the first null hypothesis was partially accepted. In general, the zirconia abutment groups exhibited lower fracture strength than the titanium abutment groups, independent of the number of supporting implants, thus the second null hypothesis was accepted.

Clinically, the extent of edentulous area dictates the number of implants that will be placed, accordingly two clinical scenarios with one- or 2- implants should not be considered as a direct treatment alternative to each other. Even though the titanium abutments have been proven to be reliable supporting two- or three-unit anterior cFDPs (26), the use of zirconia abutments supporting cFDPs has not been investigated yet, neither clinically nor mechanically. The current study suggests that the use of zirconia abutments in combination with anterior implant-supported cFDPs may be a promising alternative, however, only when supported by minimum two implants.

In this study samples exhibited 100% survival from mechanical aspect following artificial aging. No visible cracks, deformations of the restorations or implant components was detected. The findings are in agreement with the clinical studies and systematic reviews that reported 5-year outcomes of metal ceramic cFDPs supported by titanium abutments (7, 8, 10, 27). Moreover, surprisingly no screw loosening was detected, which was reported to be one of the most frequent complications for implant-supported cFDPs for 5-year follow-up (7.9 %) (10). Despite limited evidence concerning the mechanical stability of zirconia abutments supporting cFDPs, the outcomes after
artificial aging, for all the groups, interestingly even for the group that was representing more questionable clinical situation as one-implant supported zirconia abutment based one, appear to be promising.

Concerning the fracture strength of the zirconia abutments under implant-supported cFDPs, the outcomes of the current study have demonstrated similar values to the reported values of zirconia abutments under single crown implant restorations. Those studies had similar testing methods in terms of artificial aging and loading with 30° angulation to the implant axis (19). On the other hand, there is a set of studies that reported fracture strength capacities of zirconia abutments ranged between 429 and 793 N without the artificial aging (18, 28, 29). It is known that artificial aging has a significant effect on fracture resistance of zirconia abutments (30, 31). The findings of the in-vitro studies testing mechanical behavior of the zirconia abutments without the artificial aging can be expected to report higher values.

The fracture mode of the ceramic abutments in this study is similar to the findings of another in vitro study (23). It appears that the cervical aspect of the abutments represents the area of the highest torque and stress concentrations caused by levering effects. In comparison to titanium abutments, the weakest component of the system for the zirconia group was the abutment itself. Failures in the zirconia abutment groups were limited to the abutments while screws or implant replicas remained intact. Also, the load-displacement curves demonstrated no plastic deformation, which can be explained by the characteristics of the brittle materials (Figure 5B) which is in agreement with a previous study (32). Interestingly, in the present study irrelevant from the number of supporting implants in both Zr abutment groups, the fractures located above the implant shoulder leaving the implant intact. This fracture pattern is favorable because of the ease of determination of the complication and possibility of better clinical access. The results of the present study were differing from the findings of the former studies that reported all the zirconia one-piece abutments under 30° of static load demonstrated fracture below the implant shoulder (18-22). However, it is possible that there were undetected fractures below the implant shoulder in the present study, even though there was no detectable decrease in the load displacement graphics (Figure 5A and 5B). Moreover, investigations on the fracture strength of Zr abutments (18-22) has an important dissimilarity compared to the present study, which is the restoration type. Accordingly, substantial difference between the bending
forces generated by a cFDP and a single crown needs to be considered while interpreting the results.

The previous observations on the effect of aging on the mechanical properties of zirconia showed that the aging by means of cyclic loading and exposure to moisture under changing temperatures induces the low-temperature degradation of the material, which leads to spontaneous phase transformation of the zirconia crystals from tetragonal phase to the weaker monolithic phase (31, 33). Therefore, this in vitro study was designed carefully to simulate clinical conditions and all samples were exposed to artificial oral environment. The decision for the applied force orientation and direction (30° to the implant axis) was done based on previous studies in which maxillary anterior implant restorations were tested (18-22). The parameters used for cyclic loading were limited to 49 N so as to have clinically relevant approach (34-36). There are several studies reporting that mastication forces usually range between 2 and 50 N during function (35, 36). The three-dimensional load curve is programmed by the combination of the horizontal (0.5 mm) and the vertical (6 mm) motion, resulting in precisely defined vertical impact and horizontal sliding under contact. The occlusal arrangements for cantilevered FDPs, in order to minimize the bending forces, requires to avoid possible occlusal contact. However, in the present study, the loading point was at the cantilever area. The rationale behind this selection was to create the worst clinical condition. Thermo-cycling was an important part of the fatigue testing and was applied continuously and simultaneously during the dynamic loading.

As a possible limitation in this study, cantilever FDP frameworks were fabricated from a Cr-Co alloy and no veneering was done. Obviously, this selection does not comply with clinical application in the esthetic zone. Nevertheless, the rationale behind this selection was to avoid any framework failure or complication during the aging or fracture strength test of the abutments. Implant replicas instead of original implants were used in the current study. Even though the alloy used to fabricate these replicas is the same alloy used for implants from the same system, the connection between the abutment and replicas is not identical as the abutment/original implant connection. Therefore, it should be kept in mind that the failure modes may differ when the original abutments are used.

The physiological limits can be considered ranging between 50 to 400 N in the anterior zone and up to 1000N in the posterior especially for individuals with parafunctional behaviors (37-41). In relation to the reported physiological biting forces, except for group Zr-I, all the groups showed fracture strength values greater than 400 N. This clearly demonstrates that they are applicable in the anterior
zone. Within limitations of this study the authors emphasize that there is still uncertainty in predicting the performance of zirconia abutments supporting two-unit cantilevered FDPs. Therefore, further investigations are needed to verify the resistance of the zirconia abutments under cantilevered restorations, before it can be recommended for clinical application.

Conclusions

Zirconia abutments under cFDPs when supported by minimum two implants can be considered as reliable to withstand physiological occlusal forces applied in the anterior region. The fracture strength values of zirconia abutments are lower than titanium abutments irrelevant from the number of implants, hence, in situations with high load, titanium abutments should be preferred.

Acknowledgments

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Conflict of Interest

The authors declare no conflict of interest. No benefit of any kind will be received either directly or indirectly by the authors.

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### Table 1. Mean fracture to resistance values – Mean (SD)

<table>
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<th>Number of Implants</th>
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<th></th>
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<tr>
<td></td>
<td>One</td>
<td>Two</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ti</td>
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<td>664.5 (37.59)</td>
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<tr>
<td>Zr</td>
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<td>551.12 (82.19)</td>
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<td>p*</td>
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* Mann-Whitney U test. There are 8 samples in each cell.

### Table 2. Location and mode of failure after fracture resistance tests

<table>
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<tr>
<th>Groups (n=8)</th>
<th>Plastic deformation</th>
<th>Number of fractured abutment</th>
<th>Number of fractured abutment</th>
<th>Type of abutment fracture</th>
<th>The failed abutment according to the proximity to the cantilever site (1st/2nd)</th>
<th>Acrylic resin failure</th>
<th>Failure of the FDP</th>
<th>Screw failure or screw loosening</th>
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</thead>
<tbody>
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<td>Zr-I</td>
<td>0</td>
<td>8</td>
<td>0</td>
<td>7 catastrophic fracture</td>
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<td>0</td>
<td>0</td>
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<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>1 crack</td>
<td></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>Zr-II</td>
<td>0</td>
<td>8</td>
<td>0</td>
<td>8 catastrophic fractures</td>
<td>7 samples 1st abutment 1 sample both abutments</td>
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<td>0</td>
<td>0</td>
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<tr>
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Figure Legends

**Figure 1**: A representative sample for one implant groups A: one implant with Ti abutment, B: one implant with Zr abutment, C: two implants with Ti abutment, D: two implants with Zr abutment

**Figure 2**: Sample embedded in polyurethane molds at an angle of 135° to the horizontal plane to simulate clinical conditions

**Figure 3**: Static loading was applied from the same point where the fatigue loading was applied on the cantilever area, 3mm distance from the incisal edge.

**Figure 4**: Box plots of the results after the load-to-fracture test in N after the fatigue loading (n = 8). Pairwise comparisons yielded statistical differences between and within groups, before and after artificial aging (p<.005). Ti-I: single implant supporting cantilevered FDP with titanium abutments, Ti-II: two-implant supporting cantilevered FDP with titanium abutments Zr-I: single implant supporting cantilevered FDP with zirconia abutments, Zr-II: two-implant supporting cantilevered FDP with zirconia abutments

**Figure 5**: Representative graphics of load-displacement curves A: Zr-II/8 sample, B: Ti-I/2 sample

**Figure 6**: Stereomicroscope images of the samples after fracture testing A: In Ti-I group, the failure was represented by implant neck distortion, B: Zr-II specimens depicting failure of the abutment in immediate proximity to the cantilever site, C: In Zr group samples the fracture line was located at the abutment neck close to the screw level, D: No screw fracture or loosening was detected in none of the groups
Figure 5A

Figure 5B
Figure 6A

Figure 6B