Resistance to Fracture in Fixed Dental Prostheses Over Cemented and Screw-Retained Implant-Supported Zirconia Cantilevers in the Anterior Region: An In Vitro Study

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Purpose: To evaluate the resistance to fracture in cantilevered fixed dental prostheses (cFDPs) of single implant-supported zirconia cantilevers in the anterior region. Materials and Methods: Thirty-two cemented and solely screw-retained cFDPs consisting of an implant-supported crown replacing the central incisor and an attached cantilever unit in the position of the lateral incisor in the maxilla were constructed by computer-aided design (CAD) and machined by computer-aided manufacturing (CAM). For the cemented solution, a cFDP was designed on top of a customized abutment luted to an adhesive base. For screw-retained cFDPs, abutment, cement gap, and restoration of the cementable design were combined. All cFDPs were veneered manually on the facial side. Half of the samples underwent artificial aging (thermocycling and chewing simulation) before fracture tests were conducted with loads applied to the pontic either parallel to the implant axis (axial loading on the pontic) or tilted linguually by $\alpha = 45$ degrees (oblique loading on the pontic). Thus, there were eight groups differing in cFDP design, artificial aging, and load application ($n = 8$/group). If fracture ($F_{u,\text{total}}$) occurred within the implant components, the adhesive base was replaced by a cast CoCr base, and the cFDP’s fracture resistance ($F_{u,\text{cFDP}}$) was also determined. Using statistical analyses (SPSS 24, IBM), factors affecting fracture resistance were identified. Results: $F_{u,\text{total}}$ was mainly correlated to screw fractures and therefore not affected by cFDP design. Oblique loading on the pontic ($F_{u,\text{total}} = 231$ N – $352$ N), however, led to a significant ($P < .001$) decrease in ultimate load compared with axial loading on the pontic ($F_{u,\text{total}} = 611$ N – 815 N). In relation to $F_{u,\text{total}}$, $F_{u,\text{cFDP}}$ was approximately twice as high for both loading conditions. Conclusion: When relating the results to maximum occlusal forces exerted in the maxillary anterior region, single implant-supported cFDPs can be a viable restorative treatment option. Int J Oral Maxillofac Implants 2020;35:521–529. doi: 10.11607/jomi.7899

Keywords: CAD/CAM, extension bridge, fracture load, implant-supported, zirconia

Implant-supported prostheses are a reliable prosthetic treatment solution for many indications. With clinical survival rates above 90% after 10 years in situ, both the performance of implants and suprastructures is comparable to that of conventional tooth-supported fixed dental prostheses.1,2 However, in some clinical situations, the dental implant placement is challenging or even limited by medical, anatomical, and financial restrictions.3,4 Cantilevered fixed dental prostheses (cFDPs) can be a viable alternative if the adjacent teeth should not be prepared and only one implant, respectively, can be placed for the rehabilitation of the two missing teeth.5,6

However, in the anterior region, implant treatment becomes even more sophisticated due to specific aesthetic requirements. Limited gap space may not allow aesthetic and/or functional placement of the same number of implants as lost teeth; advanced soft and hard tissue surgery is frequently needed. Thus, the waiver of a second anchoring implant—for instance—and fabrication of a two-unit cFDP reduces the treatment extent and comorbidities and saves costs, and it is therefore associated with high patient satisfaction if two adjacent teeth are missing.7,8 On the other hand, the biomechanical behavior of cFDPs suggests higher rates of (technical) complications in comparison to single crowns or end-abutment FDPs. The literature9,10 suggests that the predominant complications of implant-supported cFDPs are implant fractures (rather historic), chipping, decementation, or abutment screw loosening; however, mid-term success and survival seem to be comparable with that of conventional FDPs.11

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For two-unit cFDPs in the esthetic zone, specific evidence is sparse. A recently published retrospective case series comparing two-unit implant-supported cantilevers with single crowns on adjacent implants found two-unit cFDPs to be a valid treatment alternative in the anterior region, if the space for implants is limited. To the authors’ best knowledge, there is no further information about the performance of single implant-supported zirconia cFDPs in the anterior region with a veneer solely in the facial aspect. The avoidance of a full veneer might reduce chipping, which is the most frequent complication of that kind of restoration.

The aim of this study was to evaluate the fracture load of single implant-supported zirconia cFDPs in the anterior region of the maxilla. In addition, the impact of the attachment mode, cemented vs solely screw-retained, should be analyzed. In this context, the null hypotheses were formulated that there would be no difference in fracture resistance between the test groups with regard to cFDP design, loading condition, and artificial aging.

MATERIALS AND METHODS

Simulation of the Clinical Situation
To simulate a clinical case with missing right central and lateral incisors in the maxilla, a phantom model was used (ANA-4 series, Frasaco). A polyether impression (Impregum Penta Soft, 3M Oral Care) was made, and the impression was poured with Class IV stone (Fujirock, GC Europe). The alveolar ridge was reduced in the toothless section such that its surface was located 4 mm below the soft tissue surface of the neighboring incisor. An implant with a 4.1-mm diameter and a 10-mm length (Semados SCX, BEGO Implant Systems) consisting of grade 4 titanium, yield stress given by the manufacturer with $\sigma_y = 635 \text{ MPa}$ was placed in the region of the central incisor by an experienced surgeon (A.Z.) with a tilt of 8 degrees in the anterior direction relative to the occlusal plane.

Computer-Aided Design
Afterward, the plaster model, including a scan body (PS CADP, BEGO Implant Systems) for transfer of implant position and orientation, was digitized using a three-dimensional (3D) scanner (D800, 3Shape). Based on this model scan, computer-aided design (CAD) of the two-unit cFDPs was carried out (Dental Designer, 3Shape). For half of the cFDPs ($n = 32$), an individual abutment (with titanium insert) was designed first. On top of this (screw-retained) abutment, the final cementable restoration was constructed with an anatomical design and reduced only on the facial side for later manual veneering. For the solely screw-retained variant ($n = 32$), the abutment, cement gap, and restoration of the design described earlier were merged, and the screw-channel was elongated to penetrate the occlusal surface. Thus, both designs had an identical external geometry (Fig 1). For standardized load application in the in vitro tests, the occlusal surface of the pontic contained two planar areas, the first one perpendicular to the implant axis and the second one with a 45-degree tilt around the mesiodistal axis relative to the first plane (Fig 2).

Fabrication of the Specimens
After milling (Cercon Brain Expert, Dentsply Sirona) of all abutments and cFDPs from zirconia disks (Cercon ht, Dentsply Sirona), sintering, and application of the facial veneering (Cercon Ceram Kiss, Dentsply Sirona), individual abutments and solely screw-retained cFDPs were placed on each implant after luting (Panavia21, Kuraray) it to an adhesive base (PS TiB, BEGO Implant Systems – adhesive base and screw consisting of a grade 5 titanium alloy with $\sigma_y = 875 \text{ MPa}$ [information provided by the manufacturer]) as the connective element. The prosthetic screws (Sub-TecPlus prosthetic screw, BEGO Implant Systems) were tightened with 30-Ncm torque (torque wrench, BEGO, 10–50 Ncm, REF: 55799). The implants had been embedded in...
advances in steel molds with acrylic resin (Technovit 4071, Kulzer) using a transfer key (Fig 3a) for the vertical orientation of the implant and the correct rotation of the inner hexagon around the implant axis such that after placement of the cFDP, the mesiodistal direction of the pontic was running in parallel to the mold’s edge (Fig 3b). The embedding process had to be repeated if one of these two orientation deviations from the nominal state exceeded a previously defined threshold of 5 degrees. For the cemented solution, the individual zirconia abutment and the inner crown surface of the cFDP were sandblasted (alumina, 50 µm, 1 bar) and jointed with glass ionomer cement (Ketac Cem, 3M Oral Care). To standardize the cementation process, hardening of the cement took place in a universal testing device (Zwick/Roell 2005, Zwick) with a vertical load of 200 N along the implant axis (Fig 3c).

Testing Procedures

Two different loading conditions were compared in the tests:

- Axial loading on the pontic: eccentric load application (with respect to the implant) on the pontic with the load vector parallel to the implant axis
- Oblique loading on the pontic: eccentric load application on the pontic with the load vector tilted by 45 degrees with respect to the implant axis in the oral direction

For both load cases, the load was applied 2.50 mm from the distal end of the pontic.

For each loading condition, half of the samples were exposed to artificial aging consisting of thermocycling (10,000 cycles between 6.5°C and 60°C; TC-01, SD Mechatronik) and chewing simulation (CS-04, SD Mechatronik, force magnitude 86 N) before fracture tests. During chewing simulation, samples were immersed in distilled water. Besides the use of stainless steel spheres (d = 6 mm) as antagonists, sample orientation and loading corresponded to the setup described in the following for the fracture tests. After chewing simulation, samples were inspected with regard to screw loosening and plastic deformation. Only (nearly) complete screw loosening could be checked; a slight reduction in screw prestress would not have been noticed. The check for plastic deformations was performed on a macroscopic level; ie, it was checked if the cFDP orientation exceeded the angular threshold defined for the embedding process after artificial aging.

Fracture loads were determined in a universal testing device (Zwick/Roell 2005). Loads were applied to the pontic at a crosshead speed of 0.5 mm/min oriented either parallel to the implant axis (α = 0 degrees, sample oriented horizontally) or tilted lingually by α = 45 degrees (sample tilted by 45 degrees). This resulted in eight test series differing in cFDP design, artificial aging, and load application (n = 8 for each group) (Fig 4b). Fracture of each sample was defined by a relative drop in test force below 10% of the maximum test force (Ft, total). Loads were applied at 2.5-mm distance from the distal end of the modified pontic surface. After chewing simulation, the location of the actual force application point could be checked (Fig 2).

With respect to a coordinate system (x-axis in mesial direction, y-axis in labial direction, z-axis in coronal direction) placed at the center of the implant neck, the two loading conditions, axial loading on the pontic and oblique loading on the pontic, generated the following torque vectors $\vec{T}_{\text{axial}}$ and $\vec{T}_{\text{oblique}}$ (Fig 4a).

$$\vec{T}_{\text{axial}} = \begin{pmatrix} T_{\text{axial}, x} \\ T_{\text{axial}, y} \\ T_{\text{axial}, z} \end{pmatrix} = \begin{pmatrix} -3.0 \text{ mm} \cdot F \\ -8.0 \text{ mm} \cdot F \\ 0 \text{ mm} \cdot F \end{pmatrix} = -\begin{pmatrix} 3.0 \text{ mm} \\ 8.0 \text{ mm} \end{pmatrix} F$$

$$\vec{T}_{\text{oblique}} = \begin{pmatrix} T_{\text{oblique}, x} \\ T_{\text{oblique}, y} \\ T_{\text{oblique}, z} \end{pmatrix} = \begin{pmatrix} -10.5 \text{ mm} \cdot F \\ -13.7 \text{ mm} \cdot F/\sqrt{2} \\ -8.0 \text{ mm} \cdot F/\sqrt{2} \end{pmatrix} = -\begin{pmatrix} 10.5 \text{ mm} \\ 9.7 \text{ mm} \end{pmatrix} F$$

If fracture occurred within the metallic components (implant, screw, insert), cFDPs were reattached to cast CoCr dies (Remanium 2000, Dentaurum) after removal of the inserts (Fig 1). The cast dies provided the same external geometry as an implant with attached insert, and cementation was done as described earlier for the cemented cFDPs. Subsequently, these samples could be loaded again up to the
cFDPs’ fracture resistance ($F_{\text{cFDPs}}$). Based on maximum occlusal forces reported in the literature for the anterior region, thresholds for clinical recommendation were set at 300 N for axial loading on the pontic and 150 N for oblique loading on the pontic.$^{13,14}$

To analyze fracture modes, SE imaging (JSM-6510, Jeol) was performed for representative samples of each test group.

Fig 3  Positioning of the implant of the cemented cFDP using a transfer key (a) to obtain the vertical orientation of the implant as well as later on the parallel alignment of the pontic’s mesiodistal axis with the edge of the steel molds (b). Finally, cementation of cFDP and abutment took place in a universal testing device (c).

Fig 4  (a) Scheme of the loading conditions with axial or oblique loading on the pontic as well as the most important lever arms with respect to the implant neck center and (b) the setup during fracture testing in the universal testing device.
Statistical Analysis
All statistical analyses were performed using SPSS Ver. 24 (IBM). For the descriptive presentation of the results, mean values and standard deviations were calculated. In addition, boxplot diagrams were used for visualization of fracture loads, separated for the cFDP design (solely screw-retained/cemented), the loading mode (axial/oblique), and artificial aging (with/without). The effect of these factors was investigated using an analysis of variance (ANOVA). Differences between the first and second fracture tests were analyzed with a pairwise t test. Local statistical significance was observed at $P < .05$.

RESULTS
Representative force-displacement diagrams for the different fracture test conditions are displayed in Fig 5, and all fracture resistance results are summarized in Table 1 and visualized in Figs 6a and 6b.

**Table 1** Fracture Loads of the cFDPs Determined in the In Vitro Tests for All Test Groups (n = 8/group)

<table>
<thead>
<tr>
<th>Force (N)</th>
<th>Axial loading on the pontic</th>
<th>Oblique loading on the pontic</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Solely screw-retained</td>
<td>Cemented</td>
</tr>
<tr>
<td></td>
<td>Without aging</td>
<td>With aging</td>
</tr>
<tr>
<td>$F_{u,\text{total}}$</td>
<td>611 (63)</td>
<td>613 (42)</td>
</tr>
<tr>
<td>Max</td>
<td>812 (725)</td>
<td>725 (669)</td>
</tr>
<tr>
<td>Mean</td>
<td>738 (669)</td>
<td>676 (36)</td>
</tr>
<tr>
<td>SD</td>
<td>63 (42)</td>
<td>36 (45)</td>
</tr>
</tbody>
</table>

| $F_{u,cFDP}$ | 1,152 (1,052)               | 1,179 (1,024)                | 1,196 (1,024)               | 1,298 (1,115)                |
| Max         | 1,152 (1,052)               | 1,179 (1,024)                | 1,196 (1,024)               | 1,298 (1,115)                |
| Mean        | 1,331 (1,052)               | 1,311 (1,024)                | 1,311 (1,024)               | 1,311 (1,024)                |
| SD          | 174 (94)                    | 106 (133)                    | 114 (49)                    | 36 (74)                      |

Artificial Aging
No visible damage of the ceramic structures could be detected for all samples after being exposed to artificial aging. In addition, no noticeable screw loosening was found after removal from the chewing simulator. Plastic deformation perceptible on the macroscopic level did not occur due to artificial aging.

First Test: Fracture Resistance of the Whole Assembly
Overall, the main factor influencing the fracture resistance of the whole assembly ($F_{u,\text{total}}$) in the first test was the loading condition ($P < .001$). Mean values (standard deviations) of fracture resistance of all groups loaded obliquely on the pontic were recorded at 305 N (21 N). In contrast, with 694 N (53 N), fracture resistance was more than twice as high when an axial loading on the pontic was applied. Whereas cFDP design had no significant influence ($P = .483$), artificial aging had some effect on the load-bearing capacity ($P = .038$). However, differences between mean values of any two test groups, both loaded either axially or obliquely on the pontic, never exceeded 10%.
With 231 N for oblique loading on the pontic and 611 N for axial loading on the pontic, even minimum fracture resistances clearly exceeded the thresholds (150 N and 300 N, respectively) defined for clinical recommendation. Plastic deformations within the implant components, however, start before fracture. Visible plastic deformation, based on the load-displacement curves as shown in Fig 5, were associated with load magnitudes exceeding 150 to 200 N (oblique loading on the pontic) or 400 to 500 N (axial loading on the pontic). The application of a force tilted by 45 degrees in the oral direction was therefore more critical than loading the restoration with a force parallel to the implant axis.

A much higher resilience (deflection/force) within the linear-elastic range was associated with axial loading on the pontic compared with oblique force application: whereas obliquely applied forces were associated with resiliences of 5.08 µm/N (0.90 µm/N), an axial force on the pontic led to resiliences of only 1.78 µm/N (0.33 µm/N).

**Second Test: Fracture Resistance of the cFDPs**

Two cFDPs fractured during the first test, and a further three cFDPs were damaged when removing the adhesive bases. Thus, 59 of 64 samples could be subjected to a second test. The smallest sample size (n = 6) was given for nonaged, screw-retained cFDPs tested with an axial force on the pontic, three further groups had one missing sample (n = 7), and the remaining four groups were complete (n = 8).

During the second test, fracture resistance $F_{\text{u,cFDP}}$ of nonaged cFDPs with axial loading on the pontic was found to be 1,331 N (174 N) for the solely screw-retained design, which was significantly higher ($P = .025$) than the results for the cementable design with 1,024 N (106 N). This effect was not present after aging, with both designs showing mean fracture resistances of approximately 1,100 N.

Again, the load-bearing capacity was strongly affected by the loading condition ($P < .001$). For oblique loading on the pontic, samples fractured at 657 N (69 N), which was approximately half of the fracture resistance observed for samples loaded axially on the pontic with 1,117 N (167 N). For artificial aging, no significant influence was found. When relating fracture loads of the first and second tests, $F_{\text{u,cFDP}}/F_{\text{u,total}} = 1.61$ was given for axial load application and $F_{\text{u,cFDP}}/F_{\text{u,total}} = 2.15$ for oblique load application. Thus, the fracture resistance of the cFDPs was approximately twice as high as the fracture resistance of the whole assembly. In addition, without the resilience of the intermediate abutment, cFDPs directly luted to a pin behaved much less resiliently than their respective samples in the first test (see steeper slopes of load-displacement curves displayed in Fig 5).

**Fracture Modes**

In the first test, fracture of the abutment screw was the predominant reason for failure (60/64 samples, 94%). Two samples (3%) showed a cracking of the zirconia framework (each nonaged and solely screw-retained sample loaded either axially or obliquely on the pontic). For another two samples, a rupture within the hexagonal part of the adhesive base (both failures in groups with oblique loading on the pontic and artificial aging) could be observed. Fracture surfaces of the abutment screws (Fig 7a) showed a honeycomb pattern, which is, in general, associated with predominantly tensile loading of the screws.

In the second test, 56 of 59 tested cFDPs suffered a complete fracture of the framework. The remaining three samples, belonging to groups with a cementable
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design, failed due to a crack within the separate zirconia abutment, which caused the cFDP to come loose. There were three different fracture modes: (1) 9% of the samples showed a fracture through the abutment crown (only screw-retained cFDPs) above the adhesive base location; (2) 10% showed a fracture through the connector between the abutment crown and pontic; and (3) for 76% of the cFDPs, the crack through the framework originated from the adhesive base interface. For this last, predominant fracture mode, representative fracture surfaces caused by oblique loading on the pontic are depicted for a cemented cFDP (Fig 7b) and a solely screw-retained cFDP (Fig 7c).

DISCUSSION

Based on the results, the fracture resistance of the whole assembly (first test) was significantly affected by the factors loading condition and artificial aging; ie, the respective null hypotheses had to be rejected, whereas the factor “cFDP design” had no influence as proposed by the null hypothesis. For the second test, determining the fracture resistance of the cFDPs, only the null hypothesis that artificial aging had no significant effect could be accepted.

The results of this in vitro study suggest that fracture resistance of two-unit cantilever fixed dental prostheses is favorable, even though oblique loading impinges on the pontic of the restorations. When the outcome is ranked with results of previous studies, the recently observed fracture resistance of the cantilever fixed dental prostheses is lower.15,16 Chong et al16 investigated fracture resistance of cFDP frameworks depending on length of the cantilever extension and connector diameter. Because Chong et al measured the lever arm beginning from the end of the abutment crown and the cFDP in the present study had a minimum connector cross section of 15.1 mm² (maximum dimensions: 6.8 × 3.2 mm), the test group with the 7-mm lever arm and 3 × 5 mm (rather rectangular cross section) compared best with the situation in the present study and was associated with a fracture force of 1,012 N ± 185 N. This finding was confirmed by the cFDPs’ fracture resistance found in the present investigation for axial loading on the pontic (mean fracture load $F_{\text{ucFDP}} = 1,024$ N to 1,331 N for all test groups by axial loading on the pontic). A difference was the number of implants.

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supporting the cantilever framework, in particular, two implants with Chong et al and only one implant in the present design. The restriction to only one implant may have led to much higher stresses within the implant components, most of all the screw. Therefore, the present study experienced predominantly screw fractures associated with much lower forces when testing the total assembly consisting of cFDP, adhesive base, screw, and implant (mean fracture load: \( F_{u,\text{total}} = 669 \text{ N} - 738 \text{ N} \) for all test groups by axial loading on the pontic). For oblique loading on the pontic, only studies testing the load-bearing capacity of implant abutments \(^{17,18}\) could be found. In general, as done by Foong et al,\(^ {17}\) who found a mean fracture resistance of 270 N for titanium implants, cyclic loading with a 30-degree tilt and sequentially increasing force magnitude up to fracture is applied. Park et al,\(^ {18}\) in contrast, performed such a test for several implant systems with static loading, which is closer to the test situation in the present study, and reported mean fracture loads (fracture defined by a 20% drop in test force) of 613 N to 1,255 N. The start of plastic deformations was identified at a little bit less than half of the fracture resistance. Park et al started with a lever arm of 4 mm with respect to the center of the implant neck. This lever arm increased due to deflection of the abutments during the test. Still, the lever arms existent in the present setup with a cantilever were approximately twice as long for axial loading on the pontic (bending with respect to the implant neck) and much longer for oblique loading on the pontic (bending and torsion with respect to the implant neck). In this context, the observed mean fracture resistances of approximately 700 N for axial loading on the pontic and 300 N for oblique loading on the pontic and starting plastic deformation at 400 to 500 N and 150 to 200 N, respectively, relate well to this literature.\(^ {18}\) Foong et al,\(^ {17}\) for instance, observed a similar phenomenon in titanium abutments to anchor single crowns; on the other hand, this complication was rare when one-piece zirconia abutments were used. This may be based on the probably lower fracture resistance of the one-piece zirconia abutments compared with zirconia abutments jointed with an adhesive base. In addition, the oblique load on the pontic with smaller lever arms in relation to the present load application, again, could have been causal for this difference in fracture mode. In the present study, nevertheless, the observed fracture loads and loads associated with plastic deformation were above the maximum occlusal forces expected to be exerted clinically in the anterior region. Almost all screw fractures observed in vivo show a fatigue fracture pattern arising from cyclic loading with rather high force magnitudes causing plastic deformations in contrast to the honeycomb fracture pattern originating from a forced fracture on tension. Particularly for oblique loading on the pontic, forces leading to the beginnings of plastic deformation were close to the clinical thresholds. In the literature on the clinical success of cantilever fixed dental prostheses, screw loosening and fractures are described as accounting for nearly 10% after 5 years in situ\(^ {11}\) and will be—based on the findings of the present study—the most likely clinical failure mode for single implant-supported cFDPs.

In contrast, with respect to the thresholds for clinical recommendation, the fracture resistances of the cFDPs with a monolithic design at functional surfaces of the restorations had safety factors ranging between 3 and 5. No chipping or cracks within the facial veneering could be observed after aging and fracture of the cFDP started within the zirconia, indicating that clinical complications caused by the weak veneering ceramics may be overcome as reported in another investigation.\(^ {19}\) Since screw fractures dominated all test groups, and screw loading should not have differed between the two designs, it was no surprise that the “cFDP design” factor showed no significant influence in the first test. Furthermore, the aging procedure including cyclic loading (1.2 million cycles with 86-N force magnitude) did not cause any visible remaining deformation or noticeable screw loosening, resulting in rather similar fracture resistances for aged and nonaged samples. Load-bearing capacities determined in the second test could have been diminished by previous loading in the first test and removal of the adhesive base before the second test. The use of a massive abutment to assess the cFDPs’ fracture resistance is also stated as being more critical since stress concentrations around stiff structures are in general more pronounced. Nevertheless, fracture resistance of the cFDPs themselves was approximately twice as high as that of the whole assembly, letting the authors conclude that a fracture of the ceramic components is unlikely in a clinical setting.

The situation with reduced bone height requiring a larger vertical dimension of the cFDP was deliberately chosen to simulate a rather critical clinical situation. Since, in contrast to a horizontal force component, the lever arm with respect to the implant neck is not affected by the change in vertical dimension for an axial force component, it becomes once more clear that testing restorations with oblique loads is recommended to avoid overestimation of load-bearing capacity of cFDPs in in vitro studies.

The results presented in this study refer to the BEGO Semandos SCX implant system. As seen by the study of Park et al,\(^ {18}\) load-bearing capacity of different implant systems on the dental market can vary over a huge range due to differences in geometry and material. Therefore, transfer of the findings of the present study to questions involving other implant systems has to be done with care.
CONCLUSIONS

Within the limitations of this in vitro study, all determined fracture loads and loads associated with plastic deformation were higher than with maximum occlusal forces occurring clinically in the maxillary anterior region. For the purpose of rehabilitation of two missing neighboring teeth in the maxillary anterior region, cantilever fixed dental prostheses retained by one implant might be a viable prosthetic treatment option. However, the results also indicate that dynamic occlusion should be avoided on the cantilever extension to reduce the risk of fatigue fractures. In the end, only clinical studies can confirm the promising performance with a higher grade of clinical evidence.

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