Effect of framework design on fracture load after thermal cycling and mechanical loading of implant-supported zirconia-based prostheses

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INTRODUCTION

Long-term stability of an implant-supported prosthesis is one of the most important goals of implant therapy. To evaluate long-term stability, a clinical trial is often undertaken in order to assess the applicability and performance of materials or treatments. However, clinical trials have several shortcomings, including cost, the amount of time required, difficulty in recruiting a sufficient number of participants, and variability of outcomes5. In contrast, laboratory studies require substantially less time and have important advantages, such as a reproducibility and use of standardized test variables. In laboratory studies, artificial aging with dynamic loading and thermal cycling can simulate clinical conditions. This method allows researchers to study the clinical behavior of implant-supported prostheses and yields clinically relevant findings21.

Yttria-stabilized tetragonal zirconia polycrystal (Y-TZP) has excellent biocompatibility, superior mechanical properties, and favorable esthetics3 and is therefore used as an alternative to metal alloy framework materials in implant-supported ceramic prostheses6. Although clinical trials reported successful outcomes for zirconia-based prostheses, chipping of the layering porcelain has been identified as their most important potential limitation2-7. Approaches to avoid chipping have been introduced, including customization of zirconia frameworks to provide even and appropriate porcelain thickness8, sintering of a machine-milled lithium disilicate veneer onto the zirconia framework9, and use of monolithic zirconia restorations that do not include veneer10, and layering of composite materials to the zirconia framework11,12. The use of monolithic zirconia restorations has expanded rapidly because of the reduced chipping risk of veneer, lower cost and production time, and diminished preparation depth13. However, a major drawback of monolithic zirconia restorations is that they are less translucent than silica-based ceramics, which results in undesirable esthetic outcomes10.

The mechanical properties of indirect composite materials have significantly improved, and the survivability of indirect composite restorations is now satisfactory14,15. In addition, indirect composite materials have attractive esthetic properties and a shock-absorbing effect on occlusion and are easy to fabricate, which makes them an alternative to layering porcelain materials16,17. Some studies reported that indirect composite material is promising as a veneer for zirconia-based implant-supported prostheses11,18.

Several studies investigated the effect of modifications to the framework design of implant-supported zirconia-based bilayered prostheses on fracture resistance19-21. A study reported that fracture loads of implant-supported zirconia-based bilayered prostheses significantly varied in relation to framework designs, which included uniform-thickness (UNI), anatomic (ANA), and supported anatomic (SUP) designs21. Furthermore, the fracture load of zirconia-based indirect composite–layered crowns was found to be equal to that of zirconia-based all-ceramic crowns, regardless of framework design21. However, few studies have examined the effect of zirconia framework design on long-term fracture resistance of implant-supported zirconia-based bilayered prostheses.

This study therefore investigated the effect of
three different zirconia framework designs, including UNI, ANA, and SUP designs (Fig. 1), on the results of fracture load testing of implant-supported zirconia-based bilayered prostheses after artificial aging with thermal cycling and mechanical loading. The working hypotheses to be tested were that long-term fracture loads of implant-supported zirconia-based bilayered prostheses would be associated with zirconia framework designs, and the layering materials.

MATERIALS AND METHODS

The materials assessed and their properties are summarized in Table 1. This study evaluated three framework designs for zirconia as an implant-supported prosthesis, namely, UNI, ANA, and SUP designs (Fig. 1). Each framework was layered with feldspathic porcelain or an indirect composite material.

To simulate replacement of a missing mandibular first molar, a dental implant (diameter, 5.0 mm; Implant Lab Analog, Biomet 3i, Palm Beach Gardens, FL, USA) was used. The implants were embedded in plastic specimen holders (Plastic Ring, Sankei, Tokyo, Japan), perpendicular to the horizontal plane, by using autopolymerizing acrylic resin (Technovit 4000, Heraeus Kulzer, Hanau, Germany). To simulate clinical conditions, the resin extended up to the first thread of implant bodies. This resin has an elastic modulus of 12 GPa, which approximates that of human bone (10–18 GPa)22,23).

The standardized titanium abutments (GingiHue Post WPP572G, Biomet 3i) had a platform diameter of 5.0 mm, an abutment width of 7.5 mm, a height of 7.0 mm above the shoulder, and a circular shoulder width of 0.8 mm. The titanium abutments were placed on the implants with titanium screws (Titanium Square UniScrew UNIST, Biomet 3i) by using a torque control system (Torque Driver HTD-C, Biomet 3i) at a force of 32 N. An occlusal reduction of 1.0 mm (definitive total height, 6.0 mm) was made for all abutments by using diamond rotary cutting instruments (Diamond Point FG Bur No.106RD, Shofu, Kyoto, Japan) and a laboratory air turbine (Presto Aqua, Nakanishi, Kanuma, Japan). All abutments were verified with a silicone index (Lab Silicone, Shofu), to ensure standardized form, and were polished with silicone wheels (Silicone Wheel P Type, Shofu).

In total, 66 abutment-implant complexes were randomly assigned to three groups (n=22) of zirconia framework designs, namely, UNI, ANA, and SUP designs (Fig. 1).

The UNI design had a uniformly thick (0.5 mm) zirconia framework in the axial and occlusal areas (Fig. 1a). The abutments were scanned with a measurement device (Dental Scanner SC-3, Kuraray Noritake Dental, Tokyo, Japan), and the scanned data were converted into computer-aided design (CAD) data, which were sent to a milling device to produce a uniformly thick framework of 0.5 mm.

The ANA design had a standardized layering thickness (1.2 mm) on the framework (Fig. 1b). The SUP design was similar to the ANA design but had a supportive design—a high palatal shoulder (5.0 mm) with a horizontal butt joint—for application of layering material (Fig. 1c). For the ANA and SUP designs, a full-contour waxing (Inlay Wax Medium, GC, Tokyo, Japan) was fabricated on the abutment and cut back to obtain uniform and adequate occlusal layer thickness, as would be expected for a metal-ceramic restoration. The abutment and wax pattern were double-scanned with the measurement device (Dental Scanner SC-3, Kuraray Noritake Dental), to create and merge the datasets. The merged data were then transmitted to the milling device.

All zirconia frameworks were fabricated with a commercial dental computer-aided design and computer-aided manufacturing (CAD/CAM) system (Katana, Kuraray Noritake Dental). Excluding the finish line, a cement space of 40 μm was included in all CAD data according to the manufacturer’s recommendations. The framework substructures were machine-milled by a milling device (Katana DWX-50N, Kuraray Noritake Dental) from pre-sintered zirconia blocks (Katana Zirconia HT10, Kuraray Noritake Dental) and were
Table 1 Materials assessed in the present study

<table>
<thead>
<tr>
<th>Material (Brand name)</th>
<th>Lot No.</th>
<th>Component</th>
<th>Manufacturer</th>
</tr>
</thead>
<tbody>
<tr>
<td>Implant</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Implant Lab Analog</td>
<td>1166261</td>
<td>Ti 99%</td>
<td>Biomet 3i, Palm Beach Gardens, FL, USA</td>
</tr>
<tr>
<td>Abutment</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>GiniHue Post WPP572G</td>
<td>1152336</td>
<td>Ti 99%</td>
<td>Biomet 3i</td>
</tr>
<tr>
<td>Abutment screw</td>
<td></td>
<td></td>
<td>Biomet 3i</td>
</tr>
<tr>
<td>Titanium Square UniScrew UNIST</td>
<td>119669</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Zirconia ceramic material</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Katana Zirconia HT10</td>
<td>—</td>
<td>ZrO₂ 94.4%, Y₂O₃ 5.4%</td>
<td>Kuraray Noritake Dental, Tokyo, Japan</td>
</tr>
<tr>
<td>Feldspathic porcelain for zirconia</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cerabien ZR (SBA2, A2B, E2)</td>
<td>026626, 032307, 029649</td>
<td>aluminosilicate glass, inorganic pigment</td>
<td>Kuraray Noritake Dental</td>
</tr>
<tr>
<td>Indirect composite material</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Estenia C&amp;B (OA2)</td>
<td>00048B</td>
<td>Bis-GMA, quartz powder, organic composite filler, light-polymerization catalyst</td>
<td>Kuraray Noritake Dental</td>
</tr>
<tr>
<td>Estenia C&amp;B (DA2, E2)</td>
<td>00047A, 00100A</td>
<td>urethane methacrylate monomer, glass powder, alumina micro filler, light-polymerization catalyst</td>
<td>Kuraray Noritake Dental</td>
</tr>
<tr>
<td>Porcelain liquid</td>
<td>DFOJO</td>
<td>water, propylene glycol</td>
<td>Kuraray Noritake Dental</td>
</tr>
<tr>
<td>Priming agents</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Estenia Opaque Primer</td>
<td>6C0001</td>
<td>methacrylic acid-based monomer, MDP, monomer solvent</td>
<td>Kuraray Noritake Dental</td>
</tr>
<tr>
<td>Glass-ionomer cement</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ketac Cem Easymix</td>
<td>448093</td>
<td>powder (aluminosilicate glass, maleic acrylic acid copolymer), liquid (water, tartaric acid)</td>
<td>3M ESPE, St. Paul, MN, USA</td>
</tr>
</tbody>
</table>

Bis-GMA, bisphenol-A-diglycidyl methacrylate; MDP, 10-methacryloyloxydecyl dihydrogen phosphate.

fully sintered in a heat furnace (Katana F-1, Kuraray Noritake Dental) at 1,375°C for 2 h.

The thickness of fabricated zirconia frameworks was verified with a caliper (Measuring Device 2, YDM, Tokyo, Japan) corresponding to the predetermined values for each group. Then, the frameworks were placed on the implant abutments, and adaptation was confirmed with a sharp probe (Single-end Explorer, YDM) and a polyvinyl siloxane disclosing medium (Fit Checker, GC) under observation with an optical microscope (Stemi DV4, Carl Zeiss, Jena, Germany).

The specimens in each framework design were further divided into two subgroups (n=11) according to layering material, namely, feldspathic porcelain or indirect composite material. Before layering, the layering surface of frameworks was airborne particle-abraded with 50-μm aluminum oxide particles (Hi-Aluminas, Shofu) at a pressure of 0.2 MPa and a distance of 10 mm for 20 s.

Zirconia-based all-ceramic prostheses (ZAC) ZAC specimens were layered with feldspathic porcelain (Cerabien ZR, Kuraray Noritake Dental), in accordance with the manufacturer’s recommendations. Each layering porcelain powder was mixed with the corresponding manufacturer’s liquid (Meister Liquid,
Kuraray Noritake Dental). The resulting porcelain slurry was layered on the frameworks with a special index (K854-02-000E, Tokyo Giken, Tokyo, Japan) (Fig. 2), to standardize the shape and dimensions of the prostheses, and condensed with an ultrasonic vibrator (Ceracon II, Shofu). In accordance with the exact procedure recommended by the manufacturer, thin layers of opaque porcelain (SBA2), dentin shade (A2B), and enamel shade (E2) were applied and fired in a SingleMat Porcelain Furnace (Shofu) (Table 2). In a second firing, layering porcelain was added to compensate for shrinkage during sintering. After porcelain firing, specimen thickness was confirmed with the caliper and silicone index. Then, all specimens were glazed in the furnace (Table 2). Adaptation of the finished specimens to the abutments was then confirmed, as described above.

Zirconia-based prostheses layered onto zirconia frameworks with an indirect composite material (ZIC) group

The surface of the zirconia frameworks was treated with Estenia Opaque Primer (Kuraray Noritake Dental), which contains 10-methacryloyloxydecyl dihydrogen phosphate (MDP) as a functional monomer. Two thin layers of opaque material (Estenia C&B Body Opaque OA2, Kuraray Noritake Dental) were applied to the frameworks as a high-flow bonding agent and light-cured for 90 s in a laboratory light-polymerization unit (α-light II, J. Morita) for 90 s. Glass-ionomer cement (Ketac Cem Easymix, 3M ESPE, St. Paul, MN, USA) was used to place specimens onto implant abutments, according to the manufacturer’s instructions. During cementation, a standardized load of 30 N was applied to the occlusal surface for 7 min. Excess cement was then removed with a dental explorer. All specimens were stored in distilled water at 37°C for 24 h.

The artificial aging program comprised thermal cycling and mechanical loading. A thermocycler (Thermal Shock Tester TTS–1 LM, Thomas Kagaku, Tokyo, Japan) was used to subject specimens to 10,000 cycles of thermal cycling between 5 and 55°C in water, with a 1 min dwell time for each bath. The specimens then underwent dynamic cyclic loading, namely, 1,200,000 cycles with a stainless steel ball (diameter, 6.0 mm) in a computerized masticatory simulator (K517, Tokyo Giken). A load of 49 N was applied occlusally to the central fossa of the specimens at a frequency of 1.7 Hz.

Table 2 Firing schedules for feldspathic porcelain, based on manufacturer recommendations

<table>
<thead>
<tr>
<th>Feldspathic porcelain</th>
<th>Pre-drying Temperature (°C)</th>
<th>Heating rate (°C/min)</th>
<th>Firing temperature (°C)</th>
<th>Vacuum level (mmHg)</th>
<th>Holding time (min)</th>
<th>Cooling time (min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shade Base (SBA2)</td>
<td>600</td>
<td>5</td>
<td>45</td>
<td>930</td>
<td>720</td>
<td>1</td>
</tr>
<tr>
<td>Body (A2B) and Enamel (E1)</td>
<td>600</td>
<td>8</td>
<td>45</td>
<td>935</td>
<td>720</td>
<td>1</td>
</tr>
<tr>
<td>Glaze</td>
<td>600</td>
<td>5</td>
<td>50</td>
<td>930</td>
<td>—</td>
<td>0.5</td>
</tr>
</tbody>
</table>
the specimen and stainless steel ball to equally distribute the static load (Fig. 3). Fracture resistance testing was performed at a cross-head speed of 0.5 mm/min until fracture. During testing, the compressive load was graphically recorded as a load-displacement curve. The break detector level was set at a 10% loss of maximum force\(^{27}\).

Statistical analysis was performed by using IBM SPSS Statistics version 19.0 for Windows (IBM, Armonk, NY, USA). All data were first tested for equality of variance and normality of distribution with the Levene test and Kolmogorov-Smirnov test. On the basis of the results, the data were subjected to nonparametric testing (Kruskal-Wallis test, IBM SPSS Statistics version 19.0, IBM), to evaluate differences among framework designs. Post-hoc assessment was performed by using the Mann-Whitney \(U\)-test with Bonferroni correction. Differences between the ZAC and ZIC groups for an identical framework design were analyzed with the Mann-Whitney \(U\)-test (IBM SPSS Statistics version 19.0, IBM). The significance level was set at 0.05 in all analyses.

After fracture resistance testing, specimens were observed with an optical microscope (Stemi DV4, Carl Zeiss) at 32× magnification, to determine failure mode. Failure mode was classified as layering material fracture or framework material fracture. After fracture testing, randomly selected specimens were coated with osmium by using a sputter coater (HPC-1S, Vacuum Device, Mito, Japan) for 30 s and analyzed with a scanning electron microscope (SEM; S-4300, Hitachi High Technologies, Tokyo, Japan) operating at an accelerating voltage of 15 kV.

**RESULTS**

The results of fracture load testing and statistical analysis are presented in Table 3. In the ZAC group, the median fracture loads were 4.75, 8.02, and 8.09 kN for the UNI, ANA, and SUP designs, respectively. In the ZIC group, the median fracture loads were 4.54, 5.22, and 6.99 kN for the UNI, ANA, and SUP designs, respectively. In the ZAC group, the fracture load values for the UNI design were significantly lower than those for the other two framework designs (\(p<0.05\)). In the ZIC group, fracture load testing results did not differ between the ANA design and either the UNI or SUP designs (\(p>0.05\)).

There was no significant difference between the ZAC and ZIC groups in UNI design values (\(p=0.438\)). However, fracture loads were significantly higher for the ZAC group ANA and SUP designs (\(p=0.01\) and 0.019, respectively) than for the same designs in the ZIC group.

Table 4 shows the results of fracture mode

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**Table 3** Results of fracture load (kN) testing

<table>
<thead>
<tr>
<th>Group</th>
<th>Design</th>
<th>Minimum</th>
<th>Maximum</th>
<th>Mean</th>
<th>Median</th>
<th>IQR</th>
<th>Category 1</th>
<th>Category 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>ZAC</td>
<td>UNI</td>
<td>3.34</td>
<td>6.24</td>
<td>4.71</td>
<td>4.75</td>
<td>[4.10; 5.23]</td>
<td>A</td>
<td>a</td>
</tr>
<tr>
<td></td>
<td>ANA</td>
<td>5.97</td>
<td>9.84</td>
<td>8.02</td>
<td>8.02</td>
<td>[7.28; 9.00]</td>
<td>B</td>
<td>b</td>
</tr>
<tr>
<td></td>
<td>SUP</td>
<td>7.01</td>
<td>9.55</td>
<td>8.22</td>
<td>8.09</td>
<td>[7.48; 8.94]</td>
<td>B</td>
<td>d</td>
</tr>
<tr>
<td>ZIC</td>
<td>UNI</td>
<td>3.31</td>
<td>5.31</td>
<td>4.50</td>
<td>4.54</td>
<td>[4.36; 4.71]</td>
<td>C</td>
<td>a</td>
</tr>
<tr>
<td></td>
<td>ANA</td>
<td>4.14</td>
<td>7.57</td>
<td>5.46</td>
<td>5.22</td>
<td>[4.70; 5.66]</td>
<td>C, D</td>
<td>c</td>
</tr>
</tbody>
</table>

ZAC, zirconia-based all-ceramic prostheses; ZIC, zirconia-based prostheses layered onto zirconia framework with an indirect composite material.

UNI, uniform-thickness design; ANA, anatomic design; SUP, supported anatomic design.

IQR, interquartile range.

Category 1: Identical uppercase letters indicate that the values for framework designs are not statistically different for identical layering materials (Mann-Whitney \(U\)-test with Bonferroni correction for multiple comparisons, \(p>0.05\)).

Category 2: Identical lowercase letters indicate that the values for the ZAC and ZIC groups are not statistically different for identical framework designs (Mann-Whitney \(U\)-test, \(p>0.05\)).
assessment after fracture resistance testing. For the ZAC group ANA and SUP designs, the predominant fracture mode was catastrophic fracture of the zirconia framework material (Fig. 4a). Fracture of layering material was frequently observed in the ZAC group UNI design and the ZIC group UNI (Fig. 4b) and SUP designs (Fig. 4c). Similar numbers of ZIC group ANA design specimens exhibited layering material fracture (Fig. 4d) and framework material fracture.

Figure 5 shows representative SEM images of fractured specimens after fracture resistance testing. The specimens of fracture of layering material exhibited a combination of adhesive and cohesive fracture at the layering material/zirconia framework interface (Figs. 5a and b). The SEM images also show remnants of feldspathic porcelain or indirect composite material. Representative specimens showing the interfaces of the layering porcelain and indirect composite material to zirconia frameworks are shown in Figs. 5c and d, respectively; no adhesive failure was observed at the interfaces.

Table 4 Failure mode assessment after fracture resistance testing

<table>
<thead>
<tr>
<th>Group</th>
<th>Design</th>
<th>Fracture of layering material</th>
<th>Fracture of framework material</th>
</tr>
</thead>
<tbody>
<tr>
<td>ZAC</td>
<td>UNI</td>
<td>8 (from occlusal surface to crown margin; 8)</td>
<td>3</td>
</tr>
<tr>
<td></td>
<td>ANA</td>
<td>2 (cusp; 2)</td>
<td>9</td>
</tr>
<tr>
<td></td>
<td>SUP</td>
<td>0</td>
<td>11</td>
</tr>
<tr>
<td>ZIC</td>
<td>UNI</td>
<td>7 (from occlusal surface to crown margin; 5, cusp; 2)</td>
<td>4</td>
</tr>
<tr>
<td></td>
<td>ANA</td>
<td>6 (marginal ridge; 3, cusp; 3)</td>
<td>5</td>
</tr>
<tr>
<td></td>
<td>SUP</td>
<td>9 (marginal ridge; 5, cusp; 4)</td>
<td>2</td>
</tr>
</tbody>
</table>

ZAC, zirconia-based all-ceramic prostheses; ZIC, zirconia-based prostheses layered onto zirconia framework with an indirect composite material.

UNI, uniform-thickness design; ANA: anatomic design; SUP: supported anatomic design.

Fig. 4 Photographs of representative fractured specimens after fracture resistance testing: (a) catastrophic fracture of zirconia framework material (SUP design in ZAC group), (b) fracture of layering material from occlusal area to crown margin (UNI design in ZIC group), (c) fracture of layering material at marginal ridge (SUP design in ZIC group), and (d) fracture of layering material at occlusal cusp (ANA design in ZIC group).
DISCUSSION

The present results confirm our first working hypothesis, that is, that fracture load would be affected by the framework design of implant-supported zirconia-based bilayered prostheses. The fracture loads for the UNI framework design were significantly lower than those for the SUP design in both the ZAC and ZIC groups. There was no statistically significant difference in fracture loads between ZAC and ZIC group UNI framework designs. However, fracture loads were significantly higher for ZAC group ANA and SUP designs than for the same designs in the ZIC group. Thus, the second hypothesis—that there would be differences in long-term fracture load between prostheses layered with different materials—was partially confirmed.

The framework design of implant-supported zirconia-based prostheses affected fracture load; specifically, fracture loads were significantly lower for the UNI design than for the SUP design. This finding might be attributable to the effect of the UNI and sufficient support of layering porcelain or composite material by the zirconia framework, which likely increased fracture resistance. The anatomic occlusal design of zirconia frameworks might resist lateral force caused by vertical occlusal force applied to the prostheses. These results are consistent with those of other in vitro studies, which reported higher fracture loads for anatomically designed frameworks than for frameworks with UNI\textsuperscript{20,21}. Therefore, to enhance fracture resistance of these prostheses, uniformly thick layering material and appropriate lingual support for zirconia frameworks should be provided.

In failure mode assessment, the more frequent fracture mode for the ZAC group UNI design was fracture of the layering porcelain. However, fracture of the zirconia framework was more frequent for ZAC group ANA and SUP designs. These results suggest that the presence of uniformly thick layering porcelain may help prevent fracture or chipping of layering porcelain in implant-supported zirconia-based ceramic prostheses. In the ZIC group, fracture or delamination of layering composite material was frequent for all framework designs. Fracture of layering materials is likely caused by residual stress attributable to firing or polymerization, polishing of layering materials, and inadequate bonding of layering materials to zirconia frameworks. The zirconia-layering material interface is an important limitation of these restorations\textsuperscript{28}. The bond between layering porcelain and zirconia is stronger and more durable than that between layering composite material and zirconia\textsuperscript{29}. Thus, the lower bond strength between zirconia and layering composite material is the likely reason for fracture or delamination of layering composite material in implant-supported zirconia-based composite-layered prostheses.
Fracture loads were higher for the ANA and SUP designs in the ZAC group than for those in the ZIC group. Although direct comparison of studies is limited by differences in testing protocols, a previous study reported that initial fracture resistance of zirconia-based ceramic layered prostheses was comparable to that of zirconia-based composite-layered prostheses. The discrepancy between past and present results may be due to the greater deterioration of layering composite material as compared with layering porcelain. Deterioration induced by artificial aging of layering composite material probably decreased the fracture resistance of zirconia-based composite-layered prostheses under higher occlusal loads in the ANA and SUP specimens.

The cyclic loading (1,200,000 cycles) used in this study attempted to simulate approximately 5 years of oral service. The mean fracture load of the zirconia-based prostheses exceeded 4.5 kN in all groups, which is higher than the reported physiologic maximal posterior masticatory force of 0.88 kN. Therefore, after artificial aging, zirconia-based prostheses remain strong enough to withstand clinical chewing forces in posterior applications.

A meta-analysis yielded estimated survival rates of 93.8% for conventional tooth-supported fixed dental prostheses (FDPs), 95.2% for implant-supported FDPs and 94.5% for implant-supported single restorations after an observation period of 5 years. The most frequent technical complication of implant-supported prostheses is fracture of the layering material. Although few studies have evaluated the clinical performance of implant-supported zirconia-based ceramic prostheses, existing data indicate that the fracture rate for layering porcelain is quite high (10–40%) for implant-supported zirconia-based ceramic prostheses. In addition, implant-supported FDPs had a significantly higher fracture rate (8.8% vs 2.9% for tooth-supported FDPs). This higher fracture rate is due to the lack of periodontal ligaments, which function as shock-absorbing structures, around the osseointegrated implants. In this study, the elastic modulus of resin-embedded implants was similar to that of human bone under clinical conditions. Moreover, it is important to note that the abutment materials used in laboratory studies affect fracture loads. However, this study used standardized titanium abutments as a substrate, the same as would be used under actual clinical conditions. Therefore, the experimental protocol of this study was able to accurately simulate the relevant clinical conditions.

In this study, all specimens survived simulation of the oral environment by artificial aging. The limitations of this study are that water rather than artificial saliva was used for testing and that the number of specimens tested was comparatively low. In addition, the ANA and SUP designs had standardized layering thickness of 1.2 mm, which was assessed for previous studies, and is recommended with the manufacturer’s instructions. The layering thickness for both porcelain and indirect composite materials should be standardized to compare the fracture resistance of these materials. However, the thickness of layering materials can affect the fracture resistance of zirconia-based composite-layered prostheses. Thus, future studies should investigate differences between in vitro and in vivo fracture behavior of implant-supported zirconia-based prostheses.

Within the limitations of this in vitro study, it is concluded that uniformly thick layering material and appropriate lingual support of zirconia frameworks improve fracture resistance of implant-supported zirconia-based prostheses after artificial aging. The fracture loads for ANA and SUP designs in the ZAC group were significantly higher than those in the ZIC group. All types of implant-supported zirconia-based prostheses tested after artificial aging have the potential to withstand clinical chewing forces in posterior applications.

ACKNOWLEDGMENTS

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CONFLICTS OF INTEREST

The authors state that they have no conflicts of interest.

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