In vitro comparison of the load-bearing capacity of ceramic and metal-ceramic resin-bonded fixed dental prostheses in the posterior region

Wolfgang Bömicke, Dr med dent, MSc,a Moritz Waldecker,b Johannes Krisam, MSc,c Peter Rammelsberg, Prof Dr med dent,d and Stefan Rues, Dipl Ing,e

Presupposing careful patient selection, adherence to a strict clinical protocol, and a retentive preparation design of the adhesive wing, metal-ceramic resin-bonded fixed dental prostheses (RBFDPs) retained by adhesive wings (wing-retained [WR-] RBFDPs) have achieved a 10-year survival rate, similar to that of conventional FDPs in posterior locations.1,2 Promising results have also been obtained for metal-ceramic FDPs with adhesively cemented retainers in the form of inlays (inlay-retained [IR-]RBFDPs).3 Inlays have, moreover, been evaluated for retention of metal-free RBFDPs and enable minimally invasive and complete tooth-colored restorations.4-7 Fabricated from fiber-reinforced composite and heat-pressed lithium disilicate glass ceramic, posterior IR-RBFDPs were found, however, to be particularly vulnerable to

ABSTRACT

Statement of problem. The clinical use of ceramic resin-bonded fixed dental prostheses (RBFDPs) in the posterior region is desirable for esthetic and biological reasons but has been associated with many technical problems, including fractures or chipping of the veneer. Although these problems may be overcome by using monolithic zirconia, information is lacking about the load-bearing capacity of resin-bonded monolithic zirconia restorations for replacing a molar.

Purpose. The purpose of this in vitro study was to compare the load-bearing capacity ($F_u$), the load at initial damage ($F_{1d}$), and the failure pattern of posterior RBFDPs fabricated from monolithic zirconia (MZr), veneered zirconia (VZr), and veneered cobalt-chromium (VCo).

Material and methods. For the replacement of a maxillary first molar, 4 groups (n=8) of RBFDPs differing in prosthesis material and retainer design (MZr-IR-RBFDPs, VZr-IR-RBFDPs, MZr-WR-RBFDPs, and VCo-WR-RBFDPs; IR, inlay-retained; WR, adhesive wing-retained) were fabricated with anatomic congruence of the FDP-abutment complex. The RBFDPs were subjected to thermocycling (10 000×6.5°C/60°C) and mastication simulation (30-degree oblique loading on the pontic; 1 200 000×108 N) and then loaded until failure in a universal testing machine (0.5 mm/minute). Test forces correlating with $F_u$ and $F_{1d}$ were recorded. Statistical analysis was performed by using 2-way analysis of variance (ANOVA), 2-way repeated measures ANOVA, and the Tukey honest significant differences post hoc test (2-sided $\alpha=0.05$).

Results. $F_u$ was significantly affected by retainer design ($P<0.001$) and $F_{1d}$ by both retainer design ($P<0.001$) and prosthesis material ($P<0.001$). $F_u$ was more than 2000 N for WR-RBFDPs and more than 1000 N for IR-RBFDPs (Tukey test ranking: MZr-WR-RBFDPs > VCo-WR-RBFDPs > MZr-IR-RBFDPs = VZr-IR-RBFDPs). Ceramic RBFDPs failed by complete fracture in the connector region, whereas failure of VCo-WR-RBFDPs was limited to the ceramic veneer. $F_{1d}$ was significantly lower ($P<0.004$) than $F_u$ for veneered specimens only; $F_{1d}$ started at test forces below 500 N and coincided with veneer cracking.

Conclusions. Load-bearing capacity suitable for the definitive restoration of a molar was observed for all groups. Veneered resin-bonded fixed dental prostheses, however, were susceptible to cracking of the veneer. (J Prosthet Dent 2018;119:89-96)
Clinical Implications

In a defect-related treatment approach for definitive restorations of single, posterior, bounded edentulous spaces, monolithic zirconia resin-bonded fixed dental prostheses may be a viable alternative to veneered zirconia and metal-ceramic RBFDPs, with reduced risk of technical complications.

MATERIAL AND METHODS

In this study, RBFDPs replaced a first molar (mesiodistal width, 11 mm) of a maxillary anatomic study model (ANA 4; Frasaco GmbH). For IR-RBFDPs, the abutment teeth were prepared with 2-surface inlay cavities in accordance with guidelines for the preparation of ceramic inlay-supported prostheses. Occlusal boxes were prepared by following the anatomy of the central fissure with a depth of 1.5 mm measured from the isthmus floor to the central groove and widths of 2 mm for the premolar and 3 mm for the molar abutment tooth. Proximal boxes were prepared 2 mm deeper than occlusal boxes, with axial reduction of 1.5 mm and a width of 3 mm. For WR-RBFDPs, another set of abutment teeth was prepared with a narrow chamfer finish line, proximal, palatal, and occlusal reduction of 0.5 mm, and occlusal boxes (depth, 1.5 mm; width, 1.5 mm) and proximal grooves (length, 4 mm) as retentive elements as described for posterior metal-ceramic WR-RBFDPs. In general, parallel walls were prepared by using a paralleling device with a preparation angle of 4 degrees. Internal line angles were rounded, preparations were finished by using fine grit (30–μm) diamond rotary instruments (Komet; Gebr. Brasseler GmbH & Co KG), and tooth reduction was evaluated using a silicone index (Flexitime Easy Putty; Heraeus Kulzer GmbH) and measured with a millimeter-scaled periodontal probe.

Definitive casts (GC Fujirock EP-Premium line; GC Europe N.V.) (Fig. 1) were made from polyether impressions (Impregum Penta H DuoSoft + Impregum L Duo Soft; 3M ESPE) and digitized by using a laboratory scanner (D800; 3Shape A/S).

Two designs of either IR- or WR-RBFDPs were fabricated for each model, resulting in a total of 4 study groups (n=8): monolithic zirconia IR-RBFDPs (MZr-IR-RBFDPs), veneered zirconia IR-RBFDPs (VZr-IR-RBFDPs), monolithic zirconia WR-RBFDPs (MZr-WR-RBFDPs), and metal-ceramic veneered cobalt-chromium WR-RBFDPs (VCo-WR-RBFDPs).

MZr-IR-RBFDPs and MZr-WR-RBFDPs were designed (Dental Designer; 3Shape A/S) to restore the original situation determined from a scan of the completely dentate study model with unprepared abutment teeth. In the next step, both virtual models were provided with standardized loading sites on the mesiolateral cusp (tilted by 60 degrees relative to the direction of insertion) with special 3-dimensional (3D) manipulation software (Geomagic DesignX V 5.1.0.0; 3D Systems Inc) (Fig. 2A, B) as previously described. For the veneered...
RBFDPs, the monolithic design was split into an anatomically reduced framework and a veneer structure (Fig. 2C, D). Set to a uniform thickness of 0.7 mm, the veneer covered the pontics of the VZr-IR-RBFDPs and VCo-WR-RBFDPs and, for VZr-IR-RBFDPs, also the occlusal retainer surfaces. Gingival connector embrasures were not veneered, and retained their original form. Connector cross-sections were set at 12 mm² for monolithic restorations and 9 mm² for frameworks of veneered restorations. These procedures resulted in 3D restoration files with identical position and shape of the occlusal loading site and identical external retainer geometries within corresponding subgroups for the 4 study groups. Zirconia structures were milled (Cercon Brain Expert; DeguDent GmbH) from presintered blanks of translucent 3 mol% yttria-stabilized zirconia (Cercon ht medium; DeguDent GmbH) and sintered to their final dimensions at 1500°C (Cercon Heat Plus; DeguDent GmbH).

Frameworks of metal-ceramic RBFDs were milled from a wax blank (Cercon wax; DeguDent GmbH) and cast in cobalt-chromium alloy (Remanium star; Dentaurum GmbH and Co KG). The veneer was also milled from heat-eliminatable material (VZr-IR-RBFDPs: Cercon base cast; DeguDent GmbH; VCo-WR-RBFDPs: Cercon wax; DeguDent GmbH). The veneer was attached to the frameworks and complemented anatomically with modeling wax (S-U-Ästhetikwachs-O; Schuler Dental GmbH) by using silicone templates of the basal surfaces of the respective monolithic equivalents to create similar basal geometries of veneered and monolithic RBFDs. The frameworks were overpressed (CergoPress; DeguDent GmbH) with ceramic veneer (VZr-IR-RBFDPs: Cercon ceram press; DeguDent GmbH, VCo-WR-RBFDPs: IPS InLine POM; Ivoclar Vivadent AG) and glazed in accordance with the manufacturers’ instructions.

The prepared abutment teeth were duplicated in cobalt-chromium alloy (Remanium star) and provided with standardized roots (length, 10 mm; cervical/apical cross-sectional areas, 6×10 mm to 3×5 mm). To simulate the resilience of the periodontium, the root sections were coated with heat-shrink tubing (Protect; Bahag AG) and sealed apically with 2-mm low-viscosity polysiloxane impression material (Flexitime Correct Flow; Heraeus Kulzer GmbH). The RBFDs were fitted to the metal abutments and provisionally fixed. The abutment teeth were set in aluminum blocks with autopolymerizing resin (Technovit 4071; Heraeus Kulzer GmbH) by using a paralleling device with a template resembling the negative of the pontic surface attached to the restorations to produce standardized test models.

Before definitive cementation, the metal abutments and the metal wings of VCo-WR-RBFDPs were airborne-particle abraded using 50-μm alumina particles at pressures of 0.5 MPa for the metal abutments and 0.4 MPa for the wing retainers. They were then ultrasonically cleaned in 70% isopropanol for 3 minutes and dried with oil-free air. The metal abutments were subsequently heated to 47°C in an incubator (Heraeus Function Line;
Heraeus Kulzer GmbH) such that the temperature was approximately 37°C during cementation. The zirconia bonding surfaces were conditioned in accordance with a validated procedure. This included coating with silica (Rocatec system; 3M ESPE) and applying a universal primer (Clearfil Ceramic Primer; Kuraray Noritake Dental Inc) for 60 seconds. For VZr-IR-RBFDPs, the glass ceramic part of the bonding surface of the inlay retainer was airborne abraded 50-μm alumina particles at 0.1 MPa, ultrasonically cleaned in 70% isopropanol for 3 minutes, dried with oil-free air, and subsequently protected with a silicone template. The zirconia bonding surface was then conditioned as described but with the universal primer applied to both the zirconia and the glass ceramic. The FDPs were cemented with autopolymerizing 10-methacryloyloxydecyl dihydrogen phosphate (MDP)-based resin cement (Panavia 21; Kuraray Noritake Dental Inc) under a constant axial load of 200 N (Z005; Zwick GmbH and Co KG). Excess cement was removed, and the restoration margins were coated with polymerization-activating gel (Oxyguard; Kuraray Noritake Dental Inc). After 4 minutes, the load was removed, and the test specimens were stored in the incubator at 37°C for another 6 minutes to enable complete polymerization of the cement.

After cementation, all test specimens were stored for 24 ±1 hour at 37°C and 100% relative humidity and then subjected to 10 000 thermocycles in deionized water (Thermocycler TC 1; SD Mechatronik, at bath temperatures 6.5°C and 60°C). The RBFDPs, submerged in deionized water, also underwent 1 200 000 cycles of mechanical loading (chewing simulator CS4 with additional spring-damper system; SD Mechatronik); purely vertical movement with mass m=9 kg, descending speed v=30 mm/second; spring k=43 N/mm; damper d=135 Ns/m; leading to a force of magnitude \( F_{\text{max}}=108 \) N. Each specimen was embedded in a steel mold at a tilt of 30 degrees such that the loaded cusp was oriented horizontally; the load was applied perpendicularly to the flattened mesiopalatal plane of the pontic 2 mm below the cusp by using a steel sphere 6 mm in diameter (Fig. 3).

The load-bearing capacity of each FDP was tested in a universal testing device (Z005; Zwick GmbH & Co KG) with a cross-head speed of 0.5 mm/minute by applying the load to the pontic in the same manner as that during mastication simulation. Software (testXpert II; Zwick GmbH & Co KG) was used to determine test force at prosthesis failure (\( F_u = \)load-bearing capacity), defined as the maximum test force recorded before the test force dropped to below 30% of the maximum force. During the test, RBFDPs were observed visually for structural changes, and prosthesis structure-borne sound signals were recorded with a contact microphone. After superimposition of the rectified sound signals (root-mean-square value) and the force-time diagrams, initial damage of the FDPs was noted when an interim drop in test force coincided with a structure-borne sound ≥75% of the maximum root-mean-square value recorded during the test. Accordingly, \( F_{1d} \) was defined as the test force at which initial damage was noticed. If no initial damage occurred before failure, \( F_{1d} \) was set equal to \( F_u \).

Finally, fracture patterns of failed RBFDPs were qualitatively analyzed by using an optical stereomicroscope (Stemi SR; Carl Zeiss GmbH) at ×70 magnification. \( F_u \) and \( F_{1d} \) were recorded as mean ±SD, minima, maxima, medians, and 25 and 75 percentiles. Statistical analysis (IBM SPSS Statistics v22; IBM Corp; and SAS v9.4; SAS Institute Inc) was performed by using 2-way analysis of variance (ANOVA), repeated measures 2-way ANOVA, and Tukey honest significant difference (HSD) post hoc test (2-sided \( \alpha=0.05 \)).

### RESULTS

No decementation, cracks, or fractures of the RBFDPs were observed before quasistatic failure loading. Test forces at failure (\( F_u \)) for the RBFDPs are given in Table 1. Two-way ANOVA revealed a significant effect of retainer

<table>
<thead>
<tr>
<th>Test Group</th>
<th>Force (Mean ±SD)</th>
<th>Minimum</th>
<th>Maximum</th>
<th>Median</th>
<th>25th percentile</th>
<th>75th percentile</th>
</tr>
</thead>
<tbody>
<tr>
<td>MZr-IR-RBFDP</td>
<td>1215 ±244</td>
<td>896</td>
<td>1758</td>
<td>1225</td>
<td>1048</td>
<td>1300</td>
</tr>
<tr>
<td>VZr-IR-RBFDP</td>
<td>1380 ±240</td>
<td>660</td>
<td>1793</td>
<td>1311</td>
<td>1236</td>
<td>1758</td>
</tr>
<tr>
<td>MZr-WR-RBFDP</td>
<td>2191 ±381</td>
<td>1680</td>
<td>3000</td>
<td>2103</td>
<td>1981</td>
<td>2302</td>
</tr>
<tr>
<td>VCo-WR-RBFDP</td>
<td>2084 ±229</td>
<td>1225</td>
<td>2558</td>
<td>2041</td>
<td>1975</td>
<td>2144</td>
</tr>
</tbody>
</table>

IR, inlay-retained; MZ, monolithic zirconia; RBFDP, resin-bonded fixed dental prosthesis; VCo, veneered cobalt-chromium; VZr, veneered zirconia; WR, wing-retained. Identical superscript letters indicate no statistically significant differences among test groups by Tukey test (P>0.05).
design on failure load (P < .001) but not of prosthesis material (P = .497) (Table 2). Mean $F_u$ was greater than 2000 N for WR-RBFDPs and greater than 1000 N for IR-RBFDPs; although it was significantly higher for the WR-RBFDPs than for the IR-RBFDPs (P = .001), no significant differences were found within the retainer subgroups (P = .749).

The ceramic test specimens failed cohesively, with complete fracture in the regions of one or both connectors. By contrast, frameworks of VCo-WR-RBFDPs remained intact, and failure was limited to the ceramic veneer; for these a mixed cohesive-adhesive fracture pattern was observed. Examples of typical fractures are shown in Figure 4.

Test forces at initial damage ($F_{1d}$) for the RBFDPs are given in Table 3. Two-way ANOVA revealed a significant effect for both prosthesis material (P < .001) and retainer design (P < .001) on $F_{1d}$ (Table 4). Mean $F_{1d}$ was greater than 2000 N for MZr-WR-RBFDPs and greater than 1000 N for MZr-IR-RBFDPs; it was, however, less than 1000 N for both VZr-IR-RBFDPs and VCo-WR-RBFDPs, with initial damage observed at test forces less than 500 N for the weakest test specimens. The differences between monolithic and veneered subgroups were significant (P = .021), and $F_{1d}$ was significantly higher for MZr-WR-RBFDPs than for MZr-IR-RBFDPs (P < .001); differences between the test specimens of the veneered subgroups were not significant (P = .828).

Repeated measures ANOVA revealed significant effects on the test force according to the type of damage, prosthesis material, retainer design, and the interaction of type of damage and prosthesis material (all P < .001) (Table 5); initial damage occurred at test forces significantly lower than those causing prosthesis failure for veneered test specimens (P = .004 for VZr-IR-RBFDPs and P < .001 for VCo-WR-RBFDPs) but not for monolithic test specimens (P ≥ .980).

In agreement with this, structural defects before prosthesis failure were observed for veneered test specimens only; for VZr-IR-RBFDPs and VCo-WR-RBFDPs radial cohesive veneer fracture started in the region of the gingival-palatal aspect of the pontic, in the immediate vicinity of the distal connector and propagated over the buccal surface of the pontic.

### Discussion

Fracture testing of new FDP designs can contribute to decisions about their clinical applicability, thus minimizing risks for participants in subsequent clinical trials. In this study the load-bearing capacity of monolithic and veneered zirconia inlay-retained, monolithic zirconia wing-retained, and metal-ceramic wing-retained RBFDPs was evaluated. Because ANOVA revealed significant differences among the test groups, the null hypothesis was rejected. Mean failure loads were greater than 1000 N for IR-RBFDPs and greater than 2000 N for WR-RBFDPs, which substantially exceeded the maximum occlusal forces expected in the molar region (approximately 500 N), rendering complete fracture an unlikely cause of clinical failure.

In this study, RBFDPs were fabricated with connector dimensions of at least 9 mm² for the framework of veneered restorations and 12 mm² for monolithic restorations. For 3-unit posterior zirconia-based FDPs, a minimum connector diameter of 2.7 mm has been calculated to produce frameworks with a lifetime longer than 20 years. For veneered zirconia, inlay-retained FDPs, reduction of the connector dimensions (width × height) from 3×3 mm to 3×2 mm resulted in a significant reduction of the load-bearing capacity.

For zirconia-based inlay-retained FDPs replacing a molar with minimum zirconia connector dimensions of 9 mm², mean failure loads (for complete fracture) of 1248 to 3180 N have been reported. The failure load found in this study for inlay-retained RBFDPs was, therefore, at the low end of this range. Comparability with these studies is limited, however, because of differences in the test design, particularly with regard to loading conditions. In this study, test specimens were loaded obliquely and eccentrically to replicate the most critical and therefore relevant clinical load, generating both biaxial bending and torque. This is supported by the findings of Mehl et al., who observed a reduction in the fracture resistance of inlay-retained FDPs from 1749 N to 800 N when restorations were tested eccentrically to the mesiodistal axis instead of centrically. Results from 3D finite element simulations indicating that prosthesis and abutment stresses increased by a factor of at least 1.2 to 2 when RBFDPs were loaded laterally instead of axially also support oblique loading.

Among the ceramic specimens, the higher failure loads for MZr-WR-RBFDPs might have resulted from the different connector design and more favorable stress distribution in the connector area. Failure loads for MZr-WR-RBFDPs were, moreover, comparable with those for the...
metal-ceramic control, which, with a minimum retainer wall thickness of 0.5 mm, has already been clinically successful.31 Metal abutments were used to standardize the test specimens for the anatomic variability of natural teeth.10-14 Under oblique loading conditions, extracted human or resin abutment teeth would usually fail before the restorations.36 However, the greater stiffness of metal abutments might lead to overestimation of the performance of ceramic FDPs.28 In regions with stress concentration (for example at sharp edges), greater stiffness of the substructure may result in more critical stress. Finite element analysis of this experimental design may clarify these aspects.

Abutments in the test models were resiliently attached to avoid overestimation of restoration performance.29 As shown by 3D finite element analysis, maximum tensile stresses occur in the connector region and increase with increasing degrees of freedom of the abutment teeth.7 Accordingly, ceramic FDPs in this test failed in the connector region, which is in agreement with

Table 3. Test forces at initial damage (F1d[N])

<table>
<thead>
<tr>
<th>Force</th>
<th>Test Group</th>
<th>MZr-IR-RBFDP</th>
<th>VZr-IR-RBFDP</th>
<th>MZr-WR-RBFDP</th>
<th>VCo-WR-RBFDP</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean (±SD)</td>
<td></td>
<td>1212 (±244)</td>
<td>740 (±242)</td>
<td>2042 (±460)</td>
<td>610 (±203)</td>
</tr>
<tr>
<td>Minimum</td>
<td></td>
<td>896</td>
<td>331</td>
<td>1389</td>
<td>391</td>
</tr>
<tr>
<td>Maximum</td>
<td></td>
<td>1680</td>
<td>985</td>
<td>3000</td>
<td>1046</td>
</tr>
<tr>
<td>Median</td>
<td></td>
<td>1213</td>
<td>819</td>
<td>2014</td>
<td>586</td>
</tr>
<tr>
<td>25th percentile</td>
<td></td>
<td>1048</td>
<td>543</td>
<td>1822</td>
<td>477</td>
</tr>
<tr>
<td>75th percentile</td>
<td></td>
<td>1300</td>
<td>939</td>
<td>2137</td>
<td>657</td>
</tr>
</tbody>
</table>

IR, inlay-retained; MZr, monolithic zirconia; RBFDP, resin-bonded fixed dental prosthesis; VCo, veneered cobalt-chromium; VZr, veneered zirconia; WR, wing-retained. Identical superscript letters indicate no statistically significant difference among test groups by Tukey test (P > .05).

Table 4. Effects of prosthesis material and design on test force at initial damage

<table>
<thead>
<tr>
<th>Source</th>
<th>df</th>
<th>Sum of squares</th>
<th>Mean Square</th>
<th>F</th>
<th>P</th>
</tr>
</thead>
<tbody>
<tr>
<td>Model</td>
<td>3</td>
<td>10 076 208</td>
<td>3 358 736</td>
<td>36.23</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>Error</td>
<td>28</td>
<td>2 595 939</td>
<td>92 700</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Corrected total</td>
<td>31</td>
<td>12 671 801</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Retainer design</td>
<td>1</td>
<td>2 752 737</td>
<td>2 752 737</td>
<td>29.70</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>Prosthesis material</td>
<td>2</td>
<td>9 097 957</td>
<td>4 548 979</td>
<td>49.07</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>Retainer design × prosthesis material</td>
<td>0</td>
<td>0</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
| df = degrees of freedom. Data show 2-way ANOVA for effects of prosthesis material (monolithic zirconia, veneered zirconia, or metal-ceramic) and retainer design (wing-retained or inlay-retained) on test force at initial damage (F1d[N]) of resin-bonded fixed dental prostheses. Coefficient of variation, 26.453; F1d mean, 1151; root-mean-square error, 304.466; R-squared=0.7952.
Table 5. Effect of type of damage, prosthesis material, and retainer design on test force (N)

<table>
<thead>
<tr>
<th>Source</th>
<th>df</th>
<th>Sum of Squares</th>
<th>Mean Square</th>
<th>F</th>
<th>P</th>
</tr>
</thead>
<tbody>
<tr>
<td>Model</td>
<td>7</td>
<td>21 008 555</td>
<td>3 001 222</td>
<td>29.85</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>Error</td>
<td>56</td>
<td>5 631 093</td>
<td>100 555</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Corrected total</td>
<td>63</td>
<td>26 639 648</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Damage</td>
<td>1</td>
<td>6 797 885</td>
<td>6 797 885</td>
<td>67.60</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>Prosthesis material</td>
<td>2</td>
<td>4 930 747</td>
<td>2 465 374</td>
<td>24.52</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>Prosthesis material×damage</td>
<td>2</td>
<td>4 322 642</td>
<td>2 161 321</td>
<td>21.49</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>Retainer design</td>
<td>1</td>
<td>6 520 377</td>
<td>6 520 377</td>
<td>64.84</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>Retainer design×prosthesis material</td>
<td>0</td>
<td>42 902</td>
<td>42 902</td>
<td>0.43</td>
<td>.516</td>
</tr>
<tr>
<td>Retainer design×prosthesis material×damage</td>
<td>0</td>
<td>0</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Data show repeated measures ANOVA for effect of damage (initial damage or prosthesis failure), prosthesis material (monolithic zirconia, veneered zirconia, or metal-ceramic), and retainer design (wing-retained or inlay-retained) on test force (N). Coefficient of determination, R-squared, 0.7886.

In vitro clinical aging by combined thermal and mechanical cycling has been found to substantially reduce the load to failure of feldspathic, lithium disilicate, and zirconia ceramics.22-24 For veneered zirconia FDPs, a combination of 1.2×10⁶ cycles of mechanical loading with F=50 N and 6000 thermocycles simulating the clinical service of approximately 5 years has been proposed.30 Thus, with F=108 N, clinical service exceeding 5 years might have been simulated.

Recording initial damage has been shown to provide a reliable correlation between early structural changes within the specimen and the corresponding test force.34-36 In this study, initial damage at test forces significantly lower than failure loads was noted for veneered specimens only; cracks in the veneer were observed before final fracture and at test forces below the clinically relevant threshold of 500 N for individual specimens. Similarly, during testing of the fracture resistance of veneered zirconia and metal-ceramic IR-FDPs, several authors have identified the veneer as the weakest part of the restorations.10,11,14

CONCLUSIONS

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

1. The load-bearing capacity of all prosthesis designs tested enabled their use for the definitive restoration of a single molar.
2. Veneered designs, however, may be more susceptible to such technical complications as cracking of the veneer; this occurred in the test at loads similar in magnitude to those determined clinically in the molar region.

REFERENCES


Corresponding author:
Dr Wolfgang Bömicke
Department of Prosthodontics
University Clinic Heidelberg
Im Neuenheimer Feld 400
69120 Heidelberg
GERMANY
Email: Wolfgang.Boemicke@med.uni-heidelberg.de

Acknowledgments
The authors thank Dr Ian Davies for proofreading the manuscript.


Access to The Journal of Prosthetic Dentistry Online is reserved for print subscribers!

Full-text access to The Journal of Prosthetic Dentistry Online is available for all print subscribers. To activate your individual online subscription, please visit The Journal of Prosthetic Dentistry Online. Point your browser to http://www.journals.elsevierhealth.com/periodicals/ympr/home, follow the prompts to activate online access here, and follow the instructions. To activate your account, you will need your subscriber account number, which you can find on your mailing label (note: the number of digits in your subscriber account number varies from 6 to 10). See the example below in which the subscriber account number has been circled.

Sample mailing label

This is your subscription account number

**********AUTO**SCH 3-DIGIT 001

1 V97-3 J010 12345678-9
J. H. DOE
531 MAIN ST
CENTER CITY, NY 10001-001

Personal subscriptions to The Journal of Prosthetic Dentistry Online are for individual use only and may not be transferred. Use of The Journal of Prosthetic Dentistry Online is subject to agreement to the terms and conditions as indicated online.