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Reliability and fatigue failure modes of implant-supported aluminum-oxide fixed dental prostheses

Christian F. J. Stappert, D.D.S., M.S., Ph.D.¹, Marta Baldassarri, M.S., Ph.D.², Yu Zhang, M.S., Ph.D.³, Felix Hänssler, D.D.S., Dr. med. dent.⁴, Elizabeth D. Rekow, D.D.S., Ph.D., M.B.A.⁵, and Van P. Thompson, D.D.S., Ph.D.⁶

¹Assistant Professor, Department of Periodontology and Implant Dentistry, Department of Biomaterials and Biomimetics, New York University College of Dentistry, New York, USA; Associate Professor, Department of Prosthodontics, Albert-Ludwigs-University, Freiburg, Germany

²Assistant Research Scientist, Department of Periodontology and Implant Dentistry, Department of Biomaterials and Biomimetics, New York University College of Dentistry, New York, USA

³Assistant Professor, Department of Biomaterials and Biomimetics, New York University College of Dentistry, New York, USA

⁴Visiting Scholar, Department of Periodontology and Implant Dentistry, Department of Biomaterials and Biomimetics, New York University College of Dentistry, New York, USA

⁵Professor and Senior Vice Provost for Science and Technology, New York University, New York, USA

⁶Professor and Chairman, Department of Biomaterials and Biomimetics, New York University College of Dentistry, New York, USA

Abstract

Objectives—To investigate failure modes and reliability of implant-supported aluminum-oxide three-unit fixed-dental-prostheses (FDPs) using two different veneering porcelains.

Material and methods—Thirty-six aluminum-oxide FDP-frameworks were CAD/CAM fabricated and either hand-veneered (n=18) or over-pressed (n=18). All FDPs were adhesively luted to custom-made zirconium-oxide-abutments attached to dental implant fixtures (RP-4×13mm). Specimens were stored in water prior to mechanical testing. A Step-Stress-Accelerated-Life-Test (SSALT) with three load/cycles varying profiles was developed based on initial single-load-to-failure testing. Failure was defined by veneer chipping or chipping in combination with framework fracture. SSALT was performed on each FDP inclined 30° with respect to the applied load direction. For all specimens, failure modes were analyzed using polarized-reflected-light-microscopy and scanning-electron-microscopy (SEM). Reliability was computed using Weibull analysis software (Reliasoft).

Results—The dominant failure mode for the over-pressed FDPs was buccal chipping of the porcelain in the loading area of the pontic, while hand-veneered specimens failed mainly by combined failure modes in the veneering porcelain, framework and abutments. Chipping of the porcelain occurred earlier in the over-pressed specimens (350 N/85k, load/cycles) than in the

hand-veneered (600 N/110k)(profile I). Given a mission at 300 N load and 100k or 200 K cycles the computed Weibull reliability (2-sided at 90.0 % confidence bounds) was 0.99(1/0.98) and 0.99(1/0.98) for hand-veneered FDPs, and 0.45(0.76/0.10) and 0.05(0.63/0) for over-pressed FDPs, respectively.

Conclusions—In the range of average clinical loads (300–700 N), hand-veneered aluminum-oxide FDPs showed significantly less failure by chipping of the veneer than the over-pressed. Hand-veneered FDPs under fatigue loading failed at loads 600N.

Keywords

ceramic; fatigue; fixed-partial-denture; aluminum-oxide; bridge; veneering porcelain; chewing simulation; Weibull-reliability; fracture; chipping

Introduction

In the oral environment, human teeth are subjected to occlusal forces up to 900 N in the molar region, and mastication forces can exceed 250 N to 450 N in canine and premolar region, respectively (Anderson & Pinckton 1958, Chladek et al. 2001, Gibbs et al. 1986, Tortopidis et al. 1998). For their best mechanical performance, teeth restorations require dental materials capable of resisting such high loads. In the past decades, fracture resistant metal supported restorations have been used with the disadvantage of compromised esthetics. More recently, all-ceramics restorations have been introduced for their improved esthetics realized with better translucency and for their higher biocompatibility. However, these types of restorations are less strong and tough compared to metal supported systems (Anderson & Pinckton 1958, Campbell & Sozio 1988, Mc Lean 1983).

In 1967, a high-alumina ceramic was introduced for the fabrication of a bridge pontic construction (McLean 1967). In 1989, In-Ceram Alumina was the first all-ceramic system available for single-unit restorations and three-unit anterior fixed dental prostheses (FDPs) (Bona & Kelly 2008, Haselton et al. 2000). It consisted of a porous (30% porosity) alumina scaffold infiltrated with a low viscosity glass, in order to mechanically reinforce the material. In 1991, alumina was realized in a densely sintered form, Procera (Andersson & Oden 1993), for which 99.5% of the material consists of aluminum-oxide with no amorphous glass (Wagner & Chu 1996).

Aluminum-oxide has high flexural strength (~700 MPa) and fracture toughness (~3–5 MPa m^{1/2}) (Anusavice 1996, Wagner & Chu 1996, Yilmaz et al. 2007), which are exceeded only by zirconium-oxide ceramics for dental applications (flexural strength: 900–1200 MPa; fracture toughness: 5–10 MPa m^{1/2}) (Denry & Kelly 2008, Yilmaz et al. 2007). Aluminum-oxide has good esthetics (Bonnard et al. 2001) and biocompatibility (Andersson et al. 2001, Cibirka et al. 2001, Henriksson & Jemt 2003). Nowadays, along with zirconium-oxide, this biomaterial is used for all-ceramic restoration frameworks (Della Bona & Kelly 2008, Haselton et al. 2000).

Evidence based clinical data on aluminum-oxide FDPs are only available for glass-infiltrated alumina restorations with survival rates of 90% after 5 years (Olsson et al. 2003, Vult von Steyern et al. 2001), and 83% after 10 years (Olsson et al. 2003). No long-term clinical data are available on densely sintered alumina FDPs on teeth or implants (Raigrodsky 2006). In order to provide more insights on fatigue behavior and failure mechanisms of densely sintered aluminum-oxide FDPs, it is critical to perform in vitro mechanical testing.

One of the most critical types of clinical failure for all-ceramic FDPs is chipping of the veneering porcelain (Oden et al. 1998, Sailer et al. 2007). Especially zirconium-oxide FDP frameworks are showing clinical complication rates of crazing and chipping up to 50% after two years (Fischer et al. 2009, Raigrodski et al. 2006, Sailer et al. 2007, Vult von Steyern et al. 2005). Densely sintered aluminum-oxide frameworks might offer a clinical alternative for all-ceramic FDPs. Traditionally, aluminum-oxide implant-supported FDPs are veneered with hand-built-up porcelain (sintering technique). Recently, another veneering method has been introduced by pressing the porcelain onto the ceramic core (over-pressing technique) (Baldassarri et al. 2011, Beuer et al. 2009). It is necessary to understand whether using different veneering techniques, i.e. hand-veneering (Stappert et al. 2009) versus over-pressing method, leads to a significant difference in the life time of the restoration (Stappert et al. 2009).

The aim of this study is to investigate the fatigue failure modes and reliability of implant-supported hand-veneered and over-pressed aluminum-oxide three-unit FDPs.

Materials and Methods

Specimen preparation

Thirty-six implant-supported aluminum-oxide maxillary three-unit FDPs were obtained from NobelBiocare (Gothenburg, Sweden). To standardize the specimens, a master model of a natural upper dentition was fabricated, missing the left canine and the two left premolars. Two implants (NobelReplace Straight Groovy (RP), 4×13 mm, NobelBiocare, Gothenburg, Sweden) were placed in the position of the canine and the second premolar. A wax-up of a three-unit FDP was completed, employing an ovoid-pontic first premolar. The occlusal surfaces were developed according to the opposing dentition. A silicone template of the FDP wax-up was used to make a wax-up pattern for customization of the implant abutments. Each abutment design, as well as the position of the implants in the master model, was scanned (NobelProcera Scanner, Gothenburg, Sweden). Thirty-six customized zirconium-oxide canine and thirty-six zirconium-oxide premolar abutments were CAD/CAM manufactured (NobelBiocare, Gothenburg, Sweden). Thirty-six identical FDP aluminum-oxide three-unit frameworks were CAD/CAM fabricated. The framework was designed anatomically, allowing for 1 to 2 mm veneering porcelain thickness (NobelBiocare, Gothenburg, Sweden). All frameworks were air-particle abraded on the outer surface with 100 micron Al₂O₃ at 1.0 bar pressure and 10 mm distance. A thin wash bake of the frameworks was then performed with Transpa Clear at 940°C.

Eighteen frameworks were veneered by a specialized technician with hand-built up porcelain (NobelRondo*, Gothenburg, Sweden) using a silicone template to standardize veneering porcelain dimensions. The veneering procedure involved three firings at 920°C, 910°C and 900°C, followed by two glaze firings. The veneering porcelain of the remaining eighteen specimens was over-pressed (NobelRondo Press*, *commercially available as ALLUX NR, Wieland Dental, Pforzheim, Germany); the veneering layer was waxed-up onto one FDP framework accounting for similar veneering porcelain thickness of the hand-built up FDPs. Wax sprues were added to the veneering wax-up. A master silicone casting mold was generated to cast similar veneering wax layers around the residual frameworks. The wax veneered frameworks were embedded into press cylinders. The over-pressing method consisted of one firing at 1060°C, with a preheating phase at 700°C followed by a rise temperature of 60°C/min before the veneering porcelain was pressed into the lost-wax form over the FDP frameworks. After divestment, the over-pressed FDPs were cut from the sprues with a water-cooled diamond-coated disk (Diaflex H347, Horico Dental, Berlin, Germany) and two glaze firings were performed.

To prepare the specimens for mechanical testing, the zirconium-oxide abutments were screwed onto dental titanium implant fixtures for both canine and second premolar (NobelReplace Straight Groovy, regular platform (RP), 4×13 mm, NobelBiocare, Gothenburg, Sweden) using 35 N/cm torque control. FDPs were luted adhesively (Panavia F 2.0, Kuraray-Medical-Inc) to the abutments. To control the thickness of the resin cement, each FDP was mounted on a specimen holder and a load of ~15 N (1.5 kg) was applied during the polymerization process. The titanium implants of the assembled specimens were fixed in specimen holders of the fatigue simulator using polymethylmethacrylate (PMMA) (Orthodontic Resin, DENTSPLY/Caulk, DENTSPLY Int. Inc., Milford, DE, USA). The polymethylmethacrylate embedment extended up the implant fixtures to 1.5 mm below the implant shoulder. Specimens were stored in distilled water at 37°C for at least 14 days before testing to allow for hydration of the resin cement (Silva et al. 2008).

Mechanical testing

Mechanical failure of the FDPs was defined as chipping of the porcelain (judged sufficient to require clinical replacement by two investigators) or chipping in combination with fracture of the framework.

Two hand-veneered and two over-pressed FDPs were mechanically tested using an Instron machine (Model 5566, Instron Corporation, Canton, MA). A monotonically increasing load (1 mm/min) was applied at the buccal cusp pontic area of the FDPs until failure. Specimens were inclined at 30 degrees respect to the vertical load, applied from a tungsten carbide indenter (d=6.25 mm) supplied with a spherical tip (Bonfante et al. 2010, DeLong & Douglas 1983). An aluminum foil of 0.8 mm thickness was secured between the indenter and the contact point of the FDPs pontic.

For both hand-veneered and over-pressed FDPs, static failure load values were used to develop step-stress accelerated life testing (SSALT) profiles (Abernathy 2000). This type of mechanical test is characterized by increasing values of loads and number of cycles up to failure (Nelson 1990). Three SSALT profiles were developed starting at a step load of 5 to 20% of the mean load to failure value. The FDPs (n=16 per group) were distributed as following: nine specimens were tested with profile I, four with profile II and three with profile III. However, failure of specimens during accelerated life testing due to machine error causing excessive loading of the actuator was recorded. Life time (load to cycle ratio) of these specimens was accounted for in reliability calculations up to when the machine error occurred (suspensions).

SSALT tests were performed at room temperature using a chewing simulator (ELF 3300, EnduraTEC Division of Bose, Minnetonka, MN, USA) with the same tungsten carbide indenter (d=6.25 mm) set-up in the Instron machine, vertically loading the FDPs at the buccal cusp pontic area. All specimens were inclined at 30° respect to the actuator in order to best simulate clinical conditions (DeLong & Douglas 1983). An aluminum foil of 0.8 mm thickness was secured by a thin metal band carrier between the indenter and the contact point of the FDP pontic to allow load distribution (Baldassarri et al. 2011, Yildirim M 2003). The aluminum foil was renewed every 5000 cycles to avoid foil wear out. The specimens were kept in water during testing. For each failed FDP, load and number of cycles were recorded.

Failure mode analysis

In all failed FDPs, failure modes and cracks propagation patterns (fractographic features) were observed with both polarized-reflected-light microscopy (Leica MZ APO; Leica, Bensheim, Germany) and SEM (S-3500 N, Hitachi Instruments, San Jose, CA, USA).

Additionally, epoxy resin replicas (Epofix, Struers, Copenhagen, Denmark) were created from impressions of the specimens using polyvinylsiloxane (Virtual 380, extra light body and heavy body, Ivoclar-Vivadent, Mississauga, Ontario, US). Fractographic features on fractured/chipped parts of the FDPs were also evaluated. In order to identify internal crack patterns which otherwise would have been undetected due to the opacity of the FDPs, specimens were embedded in epoxy resin (Epofix, Struers, Copenhagen, Denmark) and sectioned axially either along their mesial-distal or buccal-lingual plane (Isomet 1000, Buehler, Lake Bluff, IL).

Micro-indentation testing

Fracture toughness ($\text{MPa m}^{1/2}$) was measured in the occlusal pontic area of the veneering porcelain using a micro-indentation tester (St. Joseph, MI, USA). Three hand-veneered and three over-pressed FDPs were randomly selected. Twelve indentations were performed for each specimen with a load of 9.8 N at a dwell time of 5 s. To avoid interactions, indentations were placed at a distance at least twice the crack length from each other and from artifacts and porcelain borders (Xu et al. 1998). Fracture toughness was computed using a validated formula by Anstis et al. (Anstis et al. 1981).

Statistical analyses

Among the specimens that failed due to fatigue loading in profile I, failure loads and the number of load cycles at failure were compared between groups (hand-veneered ($n=7$) and over-pressed ($n=7$) with a t-test and a Mann-Whitney test ($\alpha=0.05$; SPSS v. 18, Chicago, IL US). Due to the limited number of FDPs tested with profiles II and III, the t-tests were only carried out for profile I specimens.

The reliability of both hand-veneered and over-pressed FDPs was determined using Weibull analysis with accelerated life testing software (Alta Pro 7, Reliasoft, Tuscon, AZ, USA) using data (failure and suspensions) from all three profiles for each test group.

A t-test data analysis ($\alpha=0.05$; SigmaPlot 11.0, Ashburn, VA, US) was performed to determine whether fracture toughness was significantly different between the hand-veneered and over-pressed porcelain. Furthermore, brittleness of veneering porcelain was calculated based on indentation results of both groups (Quinn & Quinn 1997, Rhee et al. 2001).

Results

Single load to failure testing resulted in mean load values of porcelain chipping and framework fracture of 1420 N for the hand-veneered and 1540 N for the over-pressed FDPs, respectively.

For five of the sixteen hand-veneered FDPs, a machine error occurred during accelerated life testing. Thus, for those specimens, the mechanical test was stopped before they could reach failure due to fatigue loading. In this way, the number of cycles and load at which the machine error occurred were included in the Weibull analysis as suspension data points, that is, values at which the test was suspended prior to failure due to fatigue (Abernethy 2000). Since those specimens did not reach fatigue failure, they were not included in the fracture mode analysis. As a result, eleven hand-veneered FDPs remained for failure mode observation and distributed as following: Seven were tested with profile I, two with profile II and two with profile III. For the over-pressed FDPs ($n=16$), two machine error suspensions were recorded in profile I. Therefore, failure mode analyses were restricted to fourteen over-pressed FDPs of which seven were accounted for profile I.

For specimens tested under profile I, hand-veneered FDPs failed at higher loads [$P = 807 \pm 110\text{N}$ (mean $\pm$ SD)] and after a greater number of cycles [$N = 130,000 \pm 12,000$] relative to over-pressed specimens [$579 \pm 147\text{N}$ and $108,000 \pm 15,000$ cycles]. When data were statistically analyzed, the mean ranks of the hand-veneered and over-pressed specimens for cycles to failure were 10.3 and 4.7, respectively; for load to failure, comparable ranks were 10.4 and 4.6, respectively. Failure loads and number of cycles were significantly lower in the over-pressed than the hand-veneered groups when evaluated either with a t-test ($p = 0.006$ for load and 0.009 for cycles) or a Mann Whitney test ($p = 0.007$ for load and 0.011 for cycles). For both systems, the lowest load to number of cycles ratio was associated with a less pronounced failure mode, which was buccal chipping of the veneering porcelain in the loading area of the pontic at $600\text{ N}/100\text{k}$ cycles for the hand-veneered FDPs and $350\text{ N}/85\text{k}$ cycles for the over-pressed. At the same time, the highest load to number of cycles ratio was associated with most evident failures. In these cases, a combined fracture of the pontic, premolar abutment and canine connector was observed at $950\text{ N}/150\text{k}$ for the hand-veneered FDPs and combined chipping of the porcelain veneer and fracture of both abutments at $750\text{ N}/125\text{k}$ for the over-pressed (profile I). All FDPs were analyzed as-fractured as well as sectioned (either along the mesial-distal or the buccal-lingual plane) and showed the following damage modes.

Out of the eleven hand-veneered FDPs included in the failure mode analysis, three demonstrated only veneering porcelain chipping in the buccal loading area of the pontic (Figure 1). Yet, when sectioned two of these three FDPs showed additional internal crack propagation, starting from the porcelain veneer at the gingival aspect of the pontic, propagating through the veneer and reinitiating in the aluminum-oxide core. The crack propagation continued in the aluminum-oxide framework towards the occlusal loading area and extended into the occlusal veneering porcelain (Figure 2 a, b, c). In two other FDPs, the pontic fractured from the gingival aspect of the canine connector towards the buccal loading area (Figure 3 a, b), with a crack developing discontinuously from the porcelain to the core (Figure 4). For the remaining six, either one ($n=3$) or both ($n=3$) abutments fractured and separated from the implants (Figure 5 a, b). Five of these showed additional failure modes. Three fractured along one connector from the gingival to the occlusal area. The other two fractured in the pontic from the gingival aspect of canine connector to the loading area with one of them failing also along the premolar connector.

For the over-pressed FDPs, the dominant failure mode was buccal chipping of the porcelain developing from the loading area towards the gingival aspect of the pontic. In detail, out of the fourteen over-pressed specimens included in the failure mode analysis, twelve by chipping of the veneer and two failed by initiation of fracture at the zirconium-oxide abutment level. Of the twelve FDPs demonstrating chipping of the veneer, seven showed only external failure modes, but when sectioned three presented internal crack propagation in the veneering porcelain in the gingival-buccal aspect of the pontic (delamination) (Figure 6 a, b, c) or crack initiation at the gingival aspect of the connectors (Figure 7). Out of the five FDPs with additional failure modes, one specimen chipped also in the canine and second premolar veneer and internal cracks were observed in the pontic, in the canine abutment and along the second premolar connector. One showed cracks in both abutments ($750\text{ N}/125\text{k}$ (profile I)) (Figure 8), and three in one abutment. One showed further cracks along the canine connector from the gingival towards the occlusal area and two cracks in the veneering porcelain at the gingival aspect of the pontic.

For the two over-pressed FDPs that fractured by abutment failure initiation, one failed by premolar abutment fracture resulting in fracture of the canine connector and porcelain chipping at the pontic and both abutment crowns ($850\text{ N}/45\text{k}$ (profile III)). The second FDP

failed by canine abutment fracture, resulting in framework failure and premolar implant fracture (850 N/90k (profile II)).

When fracture toughness of the porcelain was computed, values were $0.40 \pm 0.06 \text{ MPa m}^{1/2}$ for hand-veneered and $0.33 \pm 0.05 \text{ MPa m}^{1/2}$ for over-pressed FDPs ($p < 0.001$). Brittleness of the veneering porcelain was calculated as $2.7 \pm 1.0 \text{ GPa m}^{1/2}$ for the hand-veneered and $5.1 \pm 1.5 \text{ GPa m}^{1/2}$ for over-pressed FDPs ($p < 0.001$).

Weibull-stress-level probability curve analysis (2-sided at 90.0% confidence bounds) for completion of a mission of 100k and 200k cycles at a load level of 300N indicated a reliability of 0.99(1/0.98) and 0.99(1/0.98) for hand-veneered FDPs, and 0.45(0.76/0.10) and 0.05(0.63/0) for over-pressed FDPs, respectively.

Additionally, 2-parameter Weibull analysis was used to determine the distribution of failure loads for both FDP systems (Weibull 7++, Reliasoft). For hand-veneered aluminum-oxide FDPs, a characteristic strength (η) of 819.6 N (874.3/768.3, 2 sided 90% confidence bounds) was calculated and 10% unreliability was predicted to occur at a load of 619.9 N (711.1/540.4). The calculation of characteristic strength (η) of over-pressed FDPs resulted in 687.3 N (791.0/597.3) and 10% unreliability was foreseen at a load of 350.6 N (472.1/260.4) (Figure 9).

The reliability calculations demonstrated that the probability of failure by veneering porcelain chipping was significantly higher for over-pressed than for hand-veneered FDPs. Over-pressed aluminum-oxide FDPs failed by lower loads and lower cycle numbers.

Discussion

This work was conceived in order to compare the fatigue response of over-pressed and hand-veneered aluminum-oxide three-unit FDPs under simulated fatigue. The over-pressed specimens were less reliable, starting chipping in the buccal-loading area of the porcelain at lower loads and cycle numbers. Hand-veneered specimens failed differently, mainly by chipping in combination with core failure, as well as separation of the abutments from the implants, at loads/cycles significantly higher than those of the over-pressed FDPs and above the clinical average load range.

Most of the over-pressed FDPs were susceptible to contact damage, failing by buccal chipping of the porcelain from the loading area to the gingival area of the pontic. An explanation might be found in the higher residual stresses generated during shrinkage of the veneer applied all as a mass with different thicknesses over the pontic surface. On the other hand, in the hand-veneered FDPs, accumulation of high stresses in the porcelain was less likely to occur since the porcelain was applied in different phases, with a more gradual shrinkage, potentially lowering the probability of chipping. Furthermore, the hand-veneered porcelain has higher porosity than the over-pressed (Kim et al. 2006). Thus, under applied loading, the hand-veneered can absorb more energy, as the cracks must extend around the porosity they stop spreading by “disappearing” into the porcelain (Kim et al. 2006) and chipping is less likely. This assumption was confirmed by our micro-indentation results, showing that in the over-pressed FDPs veneer cracks were longer than in the hand-veneered, suggesting a relatively low fracture resistance or toughness for over-pressed porcelain compared to its hand-veneered counterpart.

When samples were sectioned, delamination of the veneer/core interface was observed in three over-pressed FDPs with failure loads starting at 500 N. Due to the higher residual stresses in the over-pressed porcelain, energy may have dissipated by propagating the crack along the veneer/core interface and eventually deflecting it back into the porcelain (Kim et

al. 2006). On the other hand, in the hand-veneered specimens, cracks developed in the porcelain veneer and reinitiated in the alumina core without significant delamination at higher loads than for the over-pressed FDPs. However, due to the elastic modulus mismatch between porcelain and aluminum-oxide, when a veneer crack reinitiated in the core often with a discontinuity at the veneer/core interface, as proposed by Kim et al. (Kim et al. 2006).

In the statistical comparison of the two groups (profile I), failure loads and number of cycles at failure were significantly lower in the over-pressed than in the hand-veneered. Moreover, core damage occurred at loads above those commonly observed clinically. Therefore, chipping of the porcelain, characteristic of the over-pressed specimens at relatively low loads (within the clinical range), is more alarming than core failure in the hand-veneered FDPs.

During loading in the buccal area of the pontic at 30 degree inclination, all FDPs were subjected to occlusal-lingual and buccal-lingual bending. With this configuration, each component/area of the specimen (i.e. framework, veneer, abutments, cement, implants) was subjected to bending, tensile, and compression stresses. One of the most critical areas was the gingival aspect of the connectors, in which high tensile stresses were concentrated (Dittmer et al. 2010, Kamposiora et al. 1996). Fractures/cracks of the pontic initiated in that area (Kou et al. 2007) and propagated either towards the occlusal side of the connectors (pure bending loading) or the loading area (rotational loading in addition to bending, due to the 30 degrees inclination of the FDP respect to the actuator). The abutments were another critical component subjected to high stresses. For both hand-veneered and over-pressed FDPs, fracture of the abutments was one of the failure modes repeatedly observed. Therefore, special attention should be given in the geometry and material design of each component of the FDPs.

In many specimens, when sectioned, internal cracks were observed, mainly on those tested with profile I (relatively small load increments with larger number of cycles per increment). Being undetectable, internal cracks are more clinically dangerous than the superficial with the risk for the FDP failure without the chance to intervene before fracture (Lawn et al. 2002). Thus, it was necessary to extend the failure mode analysis to sectioned FDPs, and not limit it only to those with external cracks and fractures.

By using step-stress accelerated life testing, the aluminum-oxide FDPs were subjected to successively higher load levels in predetermined cycle stages, and therefore time-varying stress profiles. SSALT could substantially shorten the reliability test's duration, and allowed for reliability calculations based on a load to number of cycles' ratio. Yet, the cumulative effect of the applied stresses had to be taken into account when performing the fracture analysis (Nelson 1990). The bending configuration used in this study is a plausible and perhaps valid way to measure the mechanical properties of FDPs (Seghi & Sorensen 1995). However, this configuration reflects a much more catastrophic case compared to the clinical situation (Kelly 1995), in which stresses are not limited to one point of contact loading but are more distributed and are applied to the two abutments as well. Thus, in the clinical case less bending stress is involved. Nevertheless, given the limited information on long-term clinical performance of these FDPs, a well-controlled experimental study was essential to determine whether different veneering techniques affect the mechanical performance of aluminum-oxide three-unit FDPs.

Conclusion

In this study we found that both hand-veneered and over-pressed FDPs can sustain sufficiently high loads in respect to the range of occlusal loads reported in the literature

(Chladek et al. 2001, Gibbs et al. 1986, Tortopidis et al. 1998). However, under clinical conditions, aluminum-oxide FDPs are more fatigue resistant when the porcelain is hand-built, with lower tendencies to chip relative to those with over-pressed veneers.

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References

- Abernathy, RB. The new weibull handbook. North Palm Beach; 2000.
- Anderson D, Pincton D. Masticatory stresses in normal and modified occlusion. *Journal of Dental Research*. 1958; 37:312–317. [PubMed: 13525518]
- Andersson B, Taylor A, Lang BR, Scheller H, Scharer P, Sorensen JA, Tarnow D. Alumina ceramic implant abutments used for single-tooth replacement: A prospective 1-to 3-year multicenter study. *The International Journal of Prosthodontics*. 2001; 14:432–438. [PubMed: 12066638]
- Andersson M, Oden A. A new all-ceramic crown. A dense-sintered, high-purity alumina coping with porcelain. *Acta Odontologica Scandinavica*. 1993; 51:59–64. [PubMed: 8451925]
- Anstis GR, Chantikul P, Lawn BR, Marshall DB. A critical evaluation of indentation techniques for measuring fracture toughness i. Direct crack measurements. *Journal of the American Ceramic Society*. 1981; 64:533–538.
- Anusavice, K. Phillips Science of Dental Materials. 10. WB Saunders; Philadelphia: 1996. p. 457–458.
- Baldassarri M, Zhang Y, Thompson VP, Rekow ED, Stappert CF. Reliability and failure modes of implant-supported zirconium-oxide fixed dental prostheses related to veneering techniques. *Journal of Dentistry*. 2011; 04:006. [Epub ahead of print].
- Beuer F, Edelhoff D, Gernet W, Sorensen JA. Three-year clinical prospective evaluation of zirconia-based posterior fixed dental prostheses (fdps). *Clinical Oral Investigations*. 2009; 13:445–451. [PubMed: 19169719]
- Bona AD, Kelly JR. The clinical success of all-ceramic restorations. *The Journal of the American Dental Association*. 2008; 139:8S–13. [PubMed: 18768903]
- Bonfante EA, Coelho PG, Navarro JM Jr, Pegoraro LF, Bonfante G, Thompson VP, Silva NR. Reliability and failure modes of implant-supported y-tzp and mcr three-unit bridges. *Clinical Implant Dentistry and Related Research*. 2010; 12:235–243. [PubMed: 19416277]
- Bonnard P, Hermans M, Adriaenssens P, Daelemans P, Malevez C. Anterior esthetic rehabilitation on teeth and dental implants optimized with procera technology: A case report. *Journal of Esthetic and Restorative Dentistry*. 2001; 13:163–171. [PubMed: 11499769]
- Campbell S, Sozio R. Evaluation of the fit and strength of an all-ceramic fixed partial denture. *The Journal of Prosthetic Dentistry*. 1988; 59:301–306. [PubMed: 3279185]
- Chladek W, Liski T, Karasinski A. Experimental evaluation of occlusal forces. *Acta Bioengineering and Biomechanics*. 2001; 3:25–37.
- Cibirka RM, Nelson SK, Lang BR, Rueggeberg FA. Examination of the implant-abutment interface after fatigue testing. *The Journal of Prosthetic Dentistry*. 2001; 85:268–275. [PubMed: 11264934]
- Della Bona A, Kelly JR. The clinical success of all-ceramic restorations. *The Journal of the American Dental Association*. 2008; 139(Suppl):8S–13S. [PubMed: 18768903]
- DeLong R, Douglas WH. Development of an artificial oral environment for the testing of dental restoratives: Bi-axial force and movement control. *Journal of Dental Research*. 1983; 62:32–36. [PubMed: 6571851]

- Denry I, Kelly JR. State of the art of zirconia for dental applications. *Dental Materials*. 2008; 24:299–307. [PubMed: 17659331]
- Dittmer MP, Kohorst P, Borchers L, Stiesch M. Finite element analysis to determine stress distribution in all-ceramic fpds. *Deutsche Zahnärztliche Zeitschrift*. 2010; 65:92–99.
- Fischer J, Stawarczyk B, Trottmann A, Hammerle CH. Impact of thermal properties of veneering ceramics on the fracture load of layered ce-tzp/a nanocomposite frameworks. *Dental Materials*. 2009; 25:326–330. [PubMed: 18804856]
- Gibbs CH, Mahan PE, Mauderli A, Lundeen HC, Walsh EK. Limits of human bite strength. *The Journal of Prosthetic Dentistry*. 1986; 56:226–229. [PubMed: 3463748]
- Haselton DR, Diaz-Arnold AM, Hillis SL. Clinical assessment of high-strength all-ceramic crowns. *The Journal of Prosthetic Dentistry*. 2000; 83:396–401. [PubMed: 10756288]
- Henriksson K, Jemt T. Evaluation of custom-made procera ceramic abutments for single-implant tooth replacement: A prospective 1-year follow-up study. *The International Journal of Prosthodontics*. 2003; 16:626–630. [PubMed: 14714842]
- Kamposiora P, Papavasiliou G, Bayne SC, Felton DA. Stress concentration in all-ceramic posterior fixed partial dentures. *Quintessence International*. 1996; 27:701–706. [PubMed: 9180408]
- Kelly JR. Perspectives on strength. *Dental Materials*. 1995; 11:103–110. [PubMed: 8621029]
- Kim JW, Park SH, Jung YG, Lee HS. Evaluation and control of crack propagation in dense porcelain/porous alumina layered structures for dental material applications. *Key Engineering Materials*. 2006; 317–318:457–460.
- Kou W, Kou S, Liu H, Sjogren G. Numerical modeling of the fracture process in a three-unit all-ceramic fixed partial denture. *Dental Materials*. 2007; 23:1042–1049. [PubMed: 17109952]
- Lawn BR, Deng Y, Lloyd IK, Janal MN, Rekow ED, Thompson VP. Materials design of ceramic-based layer structures for crowns. *Journal of Dental Research*. 2002; 81:433–438. [PubMed: 12097438]
- Mc Lean, J. *Dental Ceramics*. Quintessence Publishing Co Inc; Chicago: 1983. Proceedings of the first international symposium on ceramics; p. 41-58.
- McLean JW. High-alumina ceramics for bridge pontic construction. *British Dental Journal*. 1967; 123:571–577. [PubMed: 5234809]
- Nelson, WB. *Accelerated testing: Statistical models, test plans, and data analyses*. New York: John Wiley and Sons, Inc., Interscience; 1990.
- Oden A, Andersson M, Krystek-Ondracek I, Magnusson D. Five-year clinical evaluation of procera allceram crowns. *The Journal of Prosthetic Dentistry*. 1998; 80:450–456. [PubMed: 9791792]
- Olsson KG, Furst B, Andersson B, Carlsson GE. A long-term retrospective and clinical follow-up study of in-ceram alumina fpds. *The International Journal of Prosthodontics*. 2003; 16:150–156. [PubMed: 12737246]
- Quinn JB, Quinn GD. Indentation brittleness of ceramics: A fresh approach. *Journal of Materials Science*. 1997; 32:4331–4346.
- Raigrodski AJ, Chiche GJ, Potiket N, Hochstedler JL, Mohamed SE, Billiot S, Mercante DE. The efficacy of posterior three-unit zirconium-oxide-based ceramic fixed partial dental prostheses: A prospective clinical pilot study. *The Journal of Prosthetic Dentistry*. 2006; 96:237–244. [PubMed: 17052467]
- Raigrodsky A. Materials for all ceramic restorations. *Journal of Esthetic and Restorative Dentistry*. 2006; 3:117–118.
- Rhee YW, Kim HW, Deng Y, Lawn BR. Brittle fracture versus quasi plasticity in ceramics: A simple predictive index. *Journal of the American Ceramic Society*. 2001; 84:561–565.
- Sailer I, Feher A, Filser F, Gauckler LJ, Luthy H, Hammerle CH. Five-year clinical results of zirconia frameworks for posterior fixed partial dentures. *The International Journal of Prosthodontics*. 2007; 20:383–388. [PubMed: 17695869]
- Sailer I, Pjetursson BE, Zwahlen M, Hammerle CH. A systematic review of the survival and complication rates of all-ceramic and metal-ceramic reconstructions after an observation period of at least 3 years. Part ii: Fixed dental prostheses. *Clinical Oral Implants Research*. 2007; 18(Suppl 3):86–96. [PubMed: 17594373]

- Seghi RR, Sorensen JA. Relative flexural strength of six new ceramic materials. *The International Journal of Prosthodontics*. 1995; 8:239–246. [PubMed: 10348592]
- Silva NR, de Souza GM, Coelho PG, Stappert CF, Clark EA, Rekow ED, Thompson VP. Effect of water storage time and composite cement thickness on fatigue of a glass-ceramic trilayer system. *Journal of Biomedical Materials Research Part B: Applied Biomaterials*. 2008; 84:117–123.
- Stappert CF, Cabrera M, Baldassarri M, Kim B, Perez J, Pines M, Thompson VP. Failure modes of two implant-supported oxide-ceramic fixed-partial-denture systems. *Journal of Dental Research*. 2009; 88:119.
- Stappert CF, Perez J, Kim B, Cabrera M, Haenssler F, Rekow ED, Thompson VP. Reliability of implant-supported aluminum-oxide fixed-partial-dentures using different veneering porcelains. *Journal of Dental Research*. 2009; 88:1596.
- Tortopidis D, Lyons MF, Baxendale RH, Gilmour WH. The variability of bite force measurement between sessions, in different positions within the dental arch. *Journal of Oral Rehabilitation*. 1998; 25:681–686. [PubMed: 9758398]
- Vult von Steyern P, Carlson P, Nilner K. All-ceramic fixed partial dentures designed according to the dc-zirkon technique. A 2-year clinical study. *Journal of Oral Rehabilitation*. 2005; 32:180–187. [PubMed: 15707428]
- Vult von Steyern P, Jonsson O, Nilner K. Five-year evaluation of posterior all-ceramic three-unit (in-ceram) fpds. *The International Journal of Prosthodontics*. 2001; 14:379–384. [PubMed: 11508096]
- Wagner WC, Chu TM. Biaxial flexural strength and indentation fracture toughness of three new dental core ceramics. *The Journal of Prosthetic Dentistry*. 1996; 76:140–144. [PubMed: 8820804]
- Xu HH, Smith DT, Jahanmir S, Romberg E, Kelly JR, Thompson VP, Rekow ED. Indentation damage and mechanical properties of human enamel and dentin. *Journal of Dental Research*. 1998; 77:472–480. [PubMed: 9496920]
- Yildirim MFH, Marx R, Edelhoff D. In vivo fracture resistance of implant-supported all-ceramic restorations. *The Journal of Prosthetic Dentistry*. 2003; 90:325–331. [PubMed: 14564286]
- Yilmaz H, Aydin C, Gul BE. Flexural strength and fracture toughness of dental core ceramics. *The Journal of Prosthetic Dentistry*. 2007; 98:120–128. [PubMed: 17692593]

Figure 1.

a. Overview of buccal veneering porcelain chipping (VC) arresting in the apical area of the pontic of a hand-veneered FDP (8x magnification). b. Fractographic features: Origin of chipping in the load application (LA) area of the FDP pontic and incremental growth of the crack during cyclic loading towards the coronal apical region demonstrating arrest lines and twist hackle steps (black lines) (SEM, WD20.4 mm, 40x) (PA=Premolar abutment, CA=Canine abutment, white arrows = crack propagation, black arrows = crack plane turns) (In cooperation with George Quinn, NIST, USA).

Figure 2.

Hand-veneered specimen: Internal cracks in the connector/pontic area of a chipped FDP: a. Crack initiation (CI) in the gingival lingual (L) aspect of the pontic/connector (GP) area reinitiating in the aluminum-oxide core (AC) (c.) and propagating to the occlusal buccal (B) load application (LA) area (b.) (10x magnification); b. Crack proceeds from the aluminum-oxide core into the veneering porcelain to the occlusal load application area (SEM, WD34.1 mm, 300x); c. Crack propagates in the veneering porcelain towards the aluminum-oxide interface and reinitiates in the apical aspect of the aluminum-oxide core (SEM, WD33.6 mm, 300x).

Figure 3.

Hand-veneered FDP: Fracture of the pontic from the gingival aspect of the canine connector initiating in the veneering porcelain and propagating towards the load application (LA) area: a. Occlusal view: Chipping of the veneering porcelain on the buccal (B) and lingual (L) side of the FDP pontic (PO) and extended radial crack. b. Lingual view of the pontic (PL) fracture: Crack initiation at the gingival aspect (GA), crack propagation (CP) and extension to the occlusal aspect (OA) of the FDPs (12x magnification).

Figure 4.

Crack propagation (CP) initiating at the gingival aspect of a hand-veneered FDP pontic: Discontinuous crack propagation between veneering porcelain (VP) and aluminum-oxide core (AC) due to crack reinitiating in the aluminum-oxide/veneer interface (SEM, 150x, WD 33.1 mm).

Figure 5.

Fracture analyses of a premolar zirconium-oxide abutment (ZA) after fatigue loading showed two main crack propagation areas: Circumferential crack propagation related to tensile stresses at the lower contact area of the screw (S) head (upper white arrows) and cracks development under compressive and rotational stresses close to the implant (I) shoulder (lower black arrows) (SEM, WD28.9 mm, 30x) (AC = aluminum-oxide core).

Figure 6.

Over-pressed specimen: a. Internal cracks in the veneering porcelain (VP) of the pontic/connector (P/C) area of a chipped FDP (15x magnification); b. Crack initiation at the gingival aspect of the pontic/connector area (SEM, WD31.3 mm, x25), crack propagation towards and along the lingual (L) aluminum-oxide/veneer interface (b. and c.) and c. deflecting the crack back into the porcelain (SEM, WD31.3 mm, x25). (B = buccal, white arrows = crack propagation).

Figure 7.

Microscopic overview of an over-pressed FDP sectioned axially in the mesial-distal plane showing crack propagation (CP) (black arrow) initiating at the gingival aspect of the premolar connector (C) and extending through the aluminum-oxide framework into the veneering porcelain (white arrows) (14x magnification) (P = pontic, C = connector, CP= crack propagation, PA = premolar abutment).

Figure 8.

Over-pressed FDP failed by buccal porcelain chipping under high cyclic loads (750 N/125k (profile I)) demonstrating vertical crack propagation (white arrows) in both zirconium-oxide abutments on the lingual side (tensile and rotational stresses) when sectioned. a. Lingual view of the canine abutment (CA), b. lingual view of the premolar abutment (PA) (SEM, WD42.7 mm, x15).

Figure 9.

2-parameter Weibull analysis was used to determine the distribution of failure loads for hand-veneered (HV) and over-pressed (OP) aluminum-oxide FDPs (Weibull 7++, Reliasoft). The resulting Weibull modulus was $\beta = 8.0$ (11.9-5.4) for HV and $\beta = 3.3$ (4.8-2.3) for OP (2 sided 90% confidence bounds).