Influence of Preliminary Damage on the Load-Bearing Capacity of Zirconia Fixed Dental Prostheses

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Abstract

Purpose: The objective of this investigation was to evaluate the influence of differently shaped preliminary cuts in combination with artificial aging on the load-bearing capacity of four-unit zirconia fixed dental prostheses (FDPs).
Materials and Methods: Forty frameworks were fabricated from white-stage zirconia blanks (InCeram YZ, Vita) by means of a computer-aided design/computer-aided manufacturing system (Cerec inLab, Sirona). Frameworks were divided into four homogeneous groups with ten specimens each. Prior to veneering, frameworks of two groups were "damaged" by defined saw cuts of different dimensions, to simulate accidental flaws generated during shape cutting. After the veneering process, FDPs, with the exception of a control group without preliminary damage, were subjected to thermal and mechanical cycling (TMC) during 200 days storage in distilled water at 36°C. Following the aging procedure, all specimens were loaded until fracture, and forces at fracture were recorded. The statistical analysis of force at fracture data was performed using two-way ANOVA, with the level of significance chosen at 0.05.
Results: Neither type of preliminary mechanical damage significantly affected the load-bearing capacity of FDPs. In contrast, artificial aging by TMC proved to have a significant influence on the load-bearing capacity of both the undamaged and the predamaged zirconia restorations ($p < 0.001$); however, even though load-bearing capacity decreased by about 20% due to simulated aging, the FDPs still showed mean load-bearing capacities of about 1600 N.
Conclusions: The results of this study reveal that zirconia restorations have a high tolerance regarding mechanical damages. Irrespective of these findings, damage to zirconia ceramics during production or finishing should be avoided, as this may nevertheless lead to subcritical crack growth and, eventually, catastrophic failure. Furthermore, to ensure long-term clinical success, the design of zirconia restorations has to accommodate the decrease in load-bearing capacity due to TMC in the oral environment.

In recent years, all-ceramic fixed dental prostheses (FDPs) have become more widely used in the clinical practice, as a result of their good esthetic qualities, excellent biocompatibility, low plaque accumulation, and low thermal conductivity. As a consequence of the development of high-strength yttria-stabilized polycrystalline tetragonal zirconia (Y-TZP), even the heavily loaded molar region can apparently be treated with multiunit all-ceramic restorations.[1,2] Zirconia exists in three crystallographic modifications. The monoclinic phase is stable at room temperature, the tetragonal phase is stable between 1170 and 2370°C, and the cubic phase is the high temperature structure.[3] The tetragonal form can be retained in a metastable state at room temperature by doping zirconia with various oxides, in particular yttrium, which is the one commonly used for dental applications. Y-TZP owes its high strength to the so-called transformation reinforcement, a complex mechanism involving stress-induced transformation from tetragonal to monoclinic structure associated with a local 4% increase in volume.[3] This transformation takes place at a crack tip and results in compressive stresses within the matrix, which prevent further crack propagation. But despite the outstanding mechanical characteristics of Y-TZP, there are several factors within both the

Figure 1 Shaping of a zirconia framework in the connector area with a cut-off wheel.

fabricating process and the functional loading in the oral environment, which may be detrimental to the load-bearing capacity of zirconia restorations.

The strength of zirconia can be directly influenced by different surface treatments that exert different degrees and types of surface damage. These areas of surface flaws act as stress concentration sites, and even though they are microscopic in nature, they are potential origins of cracks, which could lead to subcritical crack growth and, eventually, catastrophic failure.[4] Various studies reported the detrimental influence of these surface defects on the fracture resistance of standardized zirconia specimens, whereby the reduction in strength is generally associated with the degree of surface damage.[5-8] Defects may be the result of manufacturing within the used zirconia blanks or may be inadvertently created on the material's surface by a technician during production or finishing. Sites of surface damage are quite often created in the area of the FDP's connectors when the frameworks are being shaped with a cut-off wheel (Fig 1); however, the FDPs' connectors, particularly the gingival embrasure, are the most sensitive sites for any kind of flaw. By using finite element analysis, Dittmer et al determined highest tensile stresses within a four-unit FDP at this site during functional loading.[9]

Apart from defects within the ceramic structure, strength of zirconia restorations may also be reduced by cyclic mechanical and thermal loading in the oral environment. Aqueous environments like the oral cavity generally facilitate subcritical crack growth in ceramics.[10] This subcritical crack growth occurs due to the stress-assisted reaction of water molecules with the ionic-covalent bonds at the crack tip.[11] Furthermore, hydroxyl ions penetrate into the zirconia lattice by grain boundary diffusion. These hydroxyl ions can be incorporated into the zirconia lattice by filling oxygen vacancies,[12] resulting in uncontrolled transitions from the tetragonal to the monoclinic structure of Y-TZP, accompanied by microcrack formation within the lattice.[13] Both phenomena decrease the strength of zirconia ceramics.[14,15] Furthermore, the repeated application of chewing forces contributes to a decrease in ceramic strength during service.[2,16] Strength is also degraded by the repeated thermal stressing of prosthetic restorations, which results from temperature changes caused by the consumption of hot and cold food and drink, and breathing. This thermal stressing generates tensions, which are manifested as slow subcritical crack growth, within the ceramic restorations.[17,18]

The aim of this in vitro study was to evaluate the influence of differently shaped preliminary damages (located at the gingival embrasure of the connector area) on the load-bearing capacity of four-unit zirconia FDPs. To simulate the conditions of the oral environment, FDPs were additionally subjected to artificial aging.

Materials and methods

Preparation

An upper jaw typodont plastic model (Frasaco OK 119, A-3 T, Franz Sachs & Co, Tettnang, Germany) with missing left second premolar and first molar was used. In accordance with the manufacturer's recommendations, a 1.0-mm circumferential chamfer preparation with 5° angle of convergence and an occlusal reduction of 2.0 mm was made on the first premolar and second molar to accommodate a zirconia-based all-ceramic four-unit FDP. Afterward, this situation was duplicated with an abrasion-resistant master model made of nickel-chromium alloy (Wiron 99, Bego, Bremen, Germany). Silicone impressions (Silagum, DMG, Hamburg, Germany) of this master model were made with an individual polymeric impression tray and poured in a type IV stone (Fuji Rock, GC, Leuven, Belgium). The resulting models were used as a basis for manufacturing zirconia frameworks.

Manufacture of zirconia frameworks

Forty frameworks were fabricated of presintered zirconia blanks with the help of a computer-aided design/computer-aided manufacturing (CAD/CAM) system (inLab, Sirona, Bensheim, Germany); restorations were randomly divided into four homogeneous groups of 10 specimens each (YZ_1 – YZ_4). One stone cast of the master model was optically scanned by means of active triangulation and stripe pattern projection (inEOS, Sirona). After digitization of the geometrical data, the restoration was designed by means of a CAD program (inLab 3D, Sirona) and based on the same data set, frameworks were milled (inLab, Sirona) of presintered zirconia (In-Ceram YZ Cubes, Vita, Bad Säckingen, Germany). Final sintering was performed in the furnace (inFire, Sirona) for 6 hours at 1520°C. Connector cross-sectional areas were (from mesial to distal) 12.5 mm^2, 15.6 mm^2, and 11.6 mm^2; connector width and height differed by less than 0.2 mm between frameworks. The axial wall thickness of the retainers was 0.6 mm, and the occlusal wall thickness ranged between 0.7 mm and 1.0 mm (according to position). As recommended by the manufacturer the cement space was nominally adjusted to -10 μm in the CAD/CAM system.

After fabrication, the adaptation of the frameworks to the Ni-Cr master model was conducted by an experienced technician under 4× magnification until the best possible fit was achieved. To detect the inner areas of the retainer that needed correction, a permanent marker was applied to the abutment teeth of the master model, and the frameworks were placed on the die without force. If necessary, the colored spots inside the retainers were removed by a red-ring hand piece and a fine cylindrical bur (Brasseler, Lemgo, Germany) under water cooling and light pressure.

Preliminary damage

To simulate preliminary damage of the FDPs created by a technician during shaping with a cut-off wheel, the sintered zirconia frameworks of two testing groups (YZ_3, YZ_4) were provided with a U-shaped cut at the gingival surface of the connector between the second premolar and the first molar (the presumed location of highest tensile stress during loading). This cutting was performed with an annular diamond saw (Microslice 2, Metals Research Ltd., Royston, UK) under constant water cooling and exactly defined conditions (e.g., speed of the saw, time of contact, contact pressure). The frameworks were placed in a special jig to guarantee repetitious dimensions of the damages, and all cuts were verified by means of a light optical microscope and a scanning electron microscope. To test whether the magnitude of the preliminary damage had an influence on load-bearing capacity, cuts of width 218 μm and depth 32 μm (shape 1) were made in one group (YZ_3) and cuts of width 351 μm and depth 115 μm (shape 2) were made in the other group (YZ_4) (Fig 2). In accordance with the manufacturer's recommendations, all frameworks (predamaged as well as undamaged restorations) were then subjected to regeneration firing for 15 minutes at 1000°C under atmospheric pressure.

Veneering

Veneering of the frameworks was performed with the recommended ceramic (VM9, Vita) by means of a slurry technique in accordance with the manufacturer's instructions. Homogeneous dimensioning of the veneering layer with a layer thickness between 0.5 and 1.2 mm (according to position) was guaranteed by using various silicone templates prepared in advance with a wax-up.

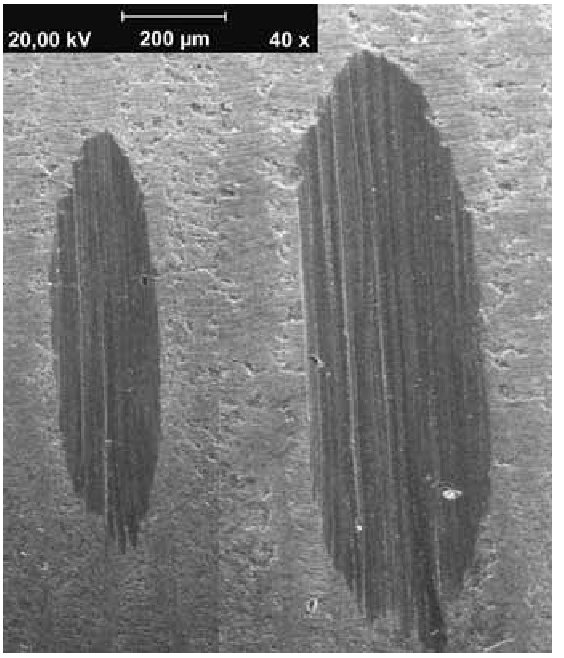

Figure 2 Scanning electron microscopy image of preliminary damages. (Left) Cut with shape 1 (YZ_3), (Right) cut with shape 2 (YZ_4).

Artificial aging

Duplicates of the prepared original abutment teeth for support of the restorations during aging and fracture testing were made of reinforced polyurethane (PUR) resin (Alpha-Die-Top, Schütz Dental, Rosbach, Germany). Physiological periodontal resilience was simulated by coating the roots of these abutments with elastic latex material (Erkoskin, Erkodent, Pfalzgrafenweiler, Germany). In the next step, all FDPs were fixed onto the PUR abutments with glass-ionomer cement (Ketac-Cem, 3M ESPE, Seefeld, Germany), and the latex-coated roots were embedded in a PUR base that extended to 3 mm below the preparation margin.

With the exception of a control group (YZ_1), the remaining undamaged (YZ_2) as well as predamaged FDPs (YZ_3, YZ_4) were subjected to thermal and mechanical cycling (TMC) during 200 days storage in distilled water at 36°C. During this period, $1{\cdot}10^4$ thermocycles between 5 and 55°C (30-second dwell time at each temperature) and $1{\cdot}10^6$ cycles of mechanical loading with 100 N as the upper limit (load frequency 2.5 Hz) were applied successively (Fig 3).

Fracture testing

After aging, all specimens were loaded to failure in a universal testing machine (Type 20 K, UTS Testsysteme, Ulm-Einsingen, Germany) at a crosshead speed of 1 mm/min with the force transferred to the occlusal connector area between the second premolar and first molar via a tungsten carbide ball (diameter 6.0 mm) and an interposed tinfoil (thickness 0.2 mm). A sudden decrease in force of more than 15 N was regarded as an indication of failure, and the maximum recorded load at that point was used as the load to failure.

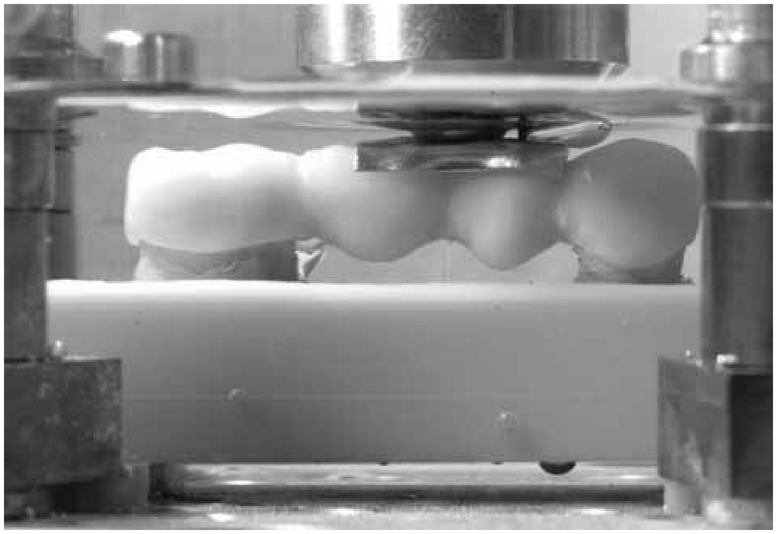

Figure 3 Cyclic loading of an fixed dental prostheses (FDP) in a chewing simulator under water storage.

Statistics

Statistical analysis was performed using SPSS for windows, version 16.0 (SPSS Software, Munich, Germany). Normal distribution of data and homogeneity of variance were checked using the Kolmogorov-Smirnov and Levene tests, respectively. Load to failure values were analyzed using two-way ANOVA, with the level of significance chosen at 0.05. The null hypotheses were that load-bearing capacity of zirconia FDPs is not influenced by preliminary damage and by artificial aging. Weibull parameters F_0 and m were determined for each test group by fitting a Weibull distribution to the respective data set. The parameter F_0 (characteristic force at failure) is associated with 63.2% probability of failure, whereas the Weibull modulus, m, is a measure of the scatter in the force at failure and of the reliability of the material investigated. The greater the value for m, the steeper is the transition from survival to failure for the probability distribution against force at failure.

Results

All restorations tested survived $1 \cdot 10^6$ cycles of mechanical loading and $1 \cdot 10^4$ thermocycles in the artificial oral environment. After the failure criterion (15 N load drop) had been fulfilled, visual inspection revealed that all FDPs failed with sudden bulk fracture of the zirconia framework. All cracks ran through the connector area between the second premolar and the first molar in a vertical direction. Detailed results of load-bearing capacity testing are shown in Table 1 and Figure 4. Fatigue by TMC proved to have a significant influence on the load-bearing capacity of the zirconia FDPs ($p < 0.001$). In comparison to the control group, the aged specimens exhibited a decrease in average force at fracture of approximately 20% (Table 1, Fig 4); however, the additional mechanical predamage of either type failed to show a statistically significant influence ($p = 0.434$) (Table 1, Fig 4).

In comparison to the corresponding forces at fracture, Weibull characteristic forces were 1.5% to 5.0% higher, but exhibited the same trends (Table 2). The application of simulated aging with mechanical and thermal loading was associated with an increase in the Weibull modulus from 9.2 (YZ_1) to 12.7 (YZ_2), although this was not statistically significant (Table 2). In comparison to the control group, the additional mechanical predamage of the zirconia frameworks in group YZ_3 caused a statistically significant increase in the mean Weibull modulus to 17.3; however, the restorations with more pronounced predamage (YZ_4) exhibited a mean Weibull modulus of only 13.4, which was not statistically significantly different from the control (Table 2).

Table 1 Loads at fracture

Group	Artificial aging	Preliminary damage	Min	Max	MD	MV	SD
YZ_1	No	No	1444.8	2337.0	2034.3	1991.4[a]	268.0
YZ_2	Yes	No	1391.0	1832.0	1593.1	1605.2[b]	133.2
YZ_3	Yes	Shape 1	1515.4	1936.8	1679.2	1685.7[b]	123.9
YZ_4	Yes	Shape 2	1395.0	1834.8	1593.0	1589.2[b]	142.0

Minimum (Min), maximum (Max), medians (MD), mean values (MV), and standard deviations (SD) are given. Superscript letters with mean values indicate statistical results. Values denoted by the same letter are not statistically significantly different.

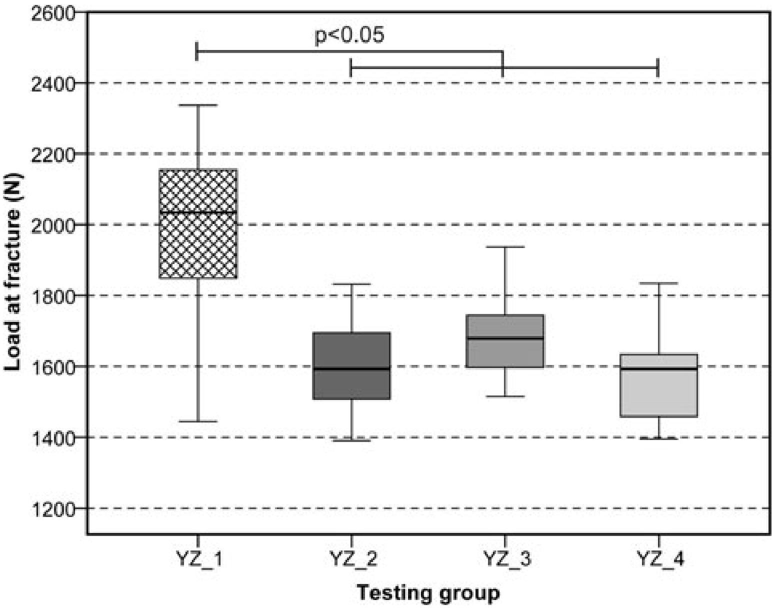

Figure 4 Box chart representing loads at fracture for each test group. Medians and quartiles are given.

Discussion

In this study, the influence of preliminary damages on the load-bearing capacity of posterior four-unit zirconia FDPs was investigated in vitro. In comparison with clinical investigations, in vitro studies are less expensive, easier to reproduce, and less prone to unpredictable influences. But to get results comparable with the in vivo situation, it is of crucial importance to design a test set-up producing a failure mode similar to that occurring clinically. Therefore, both model materials and testing conditions were chosen carefully to imitate clinical reality as faithfully as possible. The abutment teeth and their bases were made of reinforced PUR. This material has an elastic modulus somewhat lower than dentin and bone[19] but represents a better approximation of natural conditions than alloys, for example. Models made of alloys exhibit very high rigidity, so FDPs are better supported during static and cyclic loading than under the conditions in the oral cavity. This results in failure loads generally higher than those in clinical practice.[16,19]

A simple rigid support would not reproduce additional stresses caused by vertical and lateral movements during occlusal loading; therefore, a latex layer around the model abutment roots was applied to simulate periodontal resilience. At forces between 50 and 100 N in the axial direction, the abutments showed a resilience of 30 to 95 μm.[20] On the basis of published values for natural periodontal resilience,[21,22] the chosen conditions are an adequate approximation. Finally, the FDPs were cemented using a conventional luting protocol, according to clinical practice.[23]

The effect on the load-bearing capacity of zirconia restorations was investigated on two differently shaped preliminary cuts at the gingival embrasure of the middle connector. The location and the nature of the damage were deliberately chosen to resemble an unintentional flaw generated by a technician during fabrication of an FDP. Additionally, specimens were subjected to mechanical and thermal cycles in a moist environment to simulate functional loading in the oral cavity. While even a small defect introduced during the fabrication process may seem harmless at the beginning, it could have a detrimental effect on the fatigue life of zirconia restorations. With repetitive cyclic loading during mastication, small defects tend to grow in size until they reach a critical size where catastrophic failure would eventually result. Bearing in mind that subcritical crack growth starts at a threshold stress intensity factor, which for zirconia ($K_{I0} = 3.1$ MPa·m$^{1/2}$)[4] is much lower than the respective fracture toughness ($K_{Ic} = 7.4$ MPa·m$^{1/2}$),[24] the smallest

Table 2 Weibull parameters F_0 and m with their 95% confidence intervals

		95% confidence interval			95% confidence interval	
Group	F_0	Lower limit	Upper limit	m	Lower limit	Upper limit
YZ_1	2096.5	2058.5	2134.5	9.2	6.9	11.4
YZ_2	1656.7	1633.9	1679.5	12.7	9.6	15.8
YZ_3	1713.1	1691.4	1734.9	17.3	11.9	22.6
YZ_4	1632.0	1605.7	1658.2	13.4	8.8	17.9

surface defect could be large enough to act as an effective stress concentration site, thus increasing fracture probability;[25] however, neither of the two cut geometries in this study showed a significant influence on load-bearing capacity in comparison to the aged control specimens. This observation is supported by the results Kohorst et al reported in a previous study, which also evaluated restorations made of zirconia.[1] In contrast, numerous investigations dealing with preliminary surface damage of zirconia specimens have described a significant decrease in strength as a result of machining or grinding.[5,7,26-28] The reduction in strength is a function of the size and shape of surface defects.[7,8,29] Sharper and deeper defects cause greater stress concentrations and thus are more likely to act as crack initiation sites.[30] In principle, depending on the severity of surface flaws introduced, grinding of ceramics can cause two phenomena. On the one hand, it may induce residual surface compressive stress, which can considerably increase the strength of Y-TZP ceramics.[31,32] On the other hand, surface flaws may act as stress concentrators and become strength determining when they exceed the depth of the grinding-induced surface compressive layer.[33]

All restorations investigated failed with sudden bulk fracture. Cracks always ran through the connector area between the second premolar and the first molar in a vertical direction. For zirconia FDPs with the same design investigated in this study Dittmer et al reported fracture origins at the transition between framework and veneering layer at the gingival embrasure of the middle connector.[9] These fracture origins were most likely promoted by high tensile stresses within the zirconia framework. Though the likely site of highest tensile stresses during occlusal loading is located in the middle of the framework's central gingival embrasure, preliminary damage inflicted in the vicinity of this area did not create stress concentration sufficient to promote a crack origin having a significant effect on load-bearing capacity. The minor effect of the artificial damage in this investigation may be explained by the flaws being only 218 μm and 351 μm in width and 32 μm and 115 μm in depth, respectively, in the two groups. For Y-TZP specimens subjected to different surface treatments, Chevalier et al found a surface compressive layer with an average depth of up to 50 to 100 μm induced due to the transformation from tetragonal to monoclinic structure.[32] This compressive layer at the surface was thought to lead to a threshold value below which no crack growth occurs. Therefore, the damage provided to the zirconia frameworks may not have been sufficient to exceed the surface compressive layer and to act as crack initiation site. Moreover, higher stress concentrations in the regions of damage may have been prevented by the rounded inner edge design caused by the geometry of the diamond saw; however, the geometry of the saw corresponded to the geometry of a cut-off wheel routinely used by a dental technician.

A further explanation for the minor influence of the surface defects on load-bearing capacity could be the application of an additional firing step, so-called regeneration firing, after preliminary damage. Several studies have confirmed that surface treatments of Y-TZP ceramics generally trigger the tetragonal-to-monoclinic transformation.[7,8,31,34] Even if this transformation may increase the strength of Y-TZP ceramics due to residual surface compressive stresses as described above,[31,32] there is general agreement in the literature that phase transformation is associated with surface or subsurface damage, such as microcraters and grain pullout.[28,35,36] It was shown that annealing, as conducted in this study, promoted reverse transformation from the monoclinic to the tetragonal phase, thereby relaxing residual stresses within the material.[37,38] This reverse transformation may have reduced detrimental strains within the lattice of the zirconia frameworks, consequently reversing the strength-decreasing effects of the preliminary damage.

Additionally, it has to be taken into account that the scratches in the zirconia frameworks were covered by a layer of feldspathic ceramic and hence not in direct contact with the moist environment that would have otherwise facilitated further crack growth. But it is difficult to assess to what extent this protecting effect was achieved by the feldspathic ceramic used. It remains to be clarified if the veneering layer permitted sufficient water diffusion to promote zirconia degradation or if it rather acted as a protective layer, as has been reported for a combination of silica and zircon.[39]

In contrast to preliminary damage, the application of TMC in a moist environment caused a significant decrease in load-bearing capacity of approximately 20%. In a previous investigation, we found a statistically significant decrease in load-bearing capacity due to artificial aging, also with four-unit zirconia FDPs;[2] however, the mean reduction in fracture resistance was as high as 40%. Moreover, the initial mean strength of the control specimens (1991N) in this survey was considerably higher than with the FDPs evaluated in the former investigation (1525N).[2] As the dimensions of the restorations and the aging parameters were the same in both studies, the differences may have been caused by several factors: for example, composition of the zirconia materials, different sintering parameters, strength of the veneering ceramics, or the bond strength of the veneering materials to the zirconia frameworks. Furthermore, the higher load-bearing capacity of FDPs in this work may also be attributed to regeneration firing, which has not been performed in the former study. Thus, within the limitations of this study, it is unclear whether the higher fracture resistance is related to reverse transformation or the relief of any prestresses as a result of regeneration firing; further investigation is necessary.[40,41]

In contrast, Beuer et al failed to find any effect of artificial aging on load-bearing capacity in zirconia FDPs.[42] To simulate the clinical situation, they performed $1 \cdot 10^4$ thermocycles between 5 and 55°C and $1.2 \cdot 10^6$ cycles of mechanical loading with 50 N as the upper limit, conditions comparable to those of this study; however, they did not store the specimens in water over a longer period. The exposure of the restorations to an aqueous environment may decrease load-bearing capacity due to subcritical crack growth and uncontrolled transitions from the tetragonal to the monoclinic structure of Y-TZP.[12,15]

Weibull moduli in the range of 6.1 to 8.6 for zirconia-based FDPs[1,2,42] and in the range of 7 for zirconia frameworks[43] have been reported, while this study found higher Weibull moduli, in the range of 9.2 to 17.3. This might have been caused by the zirconia material used, by differences within the grinding or the sintering process, or perhaps by the regeneration firing before veneering. When the specimens were subjected to artificial aging, an increase in the Weibull modulus from 9.2 to

12.7 was found; however, this was not statistically significant. Nevertheless, this result is in accordance with the literature, where an increase of the Weibull modulus was reported after fatigue.[44] In contrast, a decrease in the Weibull modulus has been consistently reported from surface processing of zirconia with coarse milling tools.[7,34,45] However, the additional preliminary damage of the zirconia frameworks in this study resulted in a further increase in the Weibull modulus (17.3 and 13.4, for the two groups); this was statistically significant in only one testing group (shape 1). Results of the Weibull analysis indicate that the zirconia material investigated is highly reliable even after simulated aging and preliminary damage, thus qualifying this material for long-term clinical use.

Conclusions

The load-bearing capacity of zirconia FDPs tested decreased by about 20% due to simulated aging in an artificial oral environment; however, the restorations still exhibited very high mean load-bearing capacities of about 1600 N. The mean forces at fracture of various test groups were approximately 218% to 237% higher than the 500 N benchmark, which is considered to be the lower limit of static load-bearing capacity for clinically acceptable FDPs in the posterior region, taking into account cyclic fatigue loading and stress corrosion fatigue caused by the oral environment. The preliminary damage of the zirconia frameworks-–provided to simulate flaws created by a technician during shaping with a cut-off wheel—did not have a detrimental effect on the load-bearing capacity. These results demonstrate that Y-TZP restorations have a high tolerance for mechanical damages. Irrespective of these findings, damage of zirconia ceramics during production or finishing should be avoided, as this nevertheless could lead to subcritical crack growth and, eventually, catastrophic failure. Furthermore, to ensure clinical long-term success, the design of zirconia restorations has to accommodate the decrease in load-bearing capacity due to TMC in the oral environment.

References

1. Kohorst P, Herzog TJ, Borchers L, et al: Load-bearing capacity of all-ceramic posterior four-unit fixed partial dentures with different zirconia frameworks. Eur J Oral Sci 2007;115:161-166
2. Kohorst P, Dittmer MP, Borchers L, et al: Influence of cyclic fatigue in water on the load-bearing capacity of dental bridges made of zirconia. Acta Biomater 2008;4:1440-1447
3. Hannink RHJ, Kelly PM, Muddle BC: Transformation toughening in zirconia-containing ceramics. J Am Ceram Soc 2000;83:461-487
4. Aboushelib MN, de Jager N, Kleverlaan CJ, et al: Effect of loading method on the fracture mechanics of two layered all-ceramic restorative systems. Dent Mater 2007;23:952-959
5. Lee SK, Tandon R, Readey MJ, et al: Scratch damage in zirconia ceramics. J Am Ceram Soc 2000;83:1428-1432
6. Zhang Y, Lawn B: Long-term strength of ceramics for biomedical applications. J Biomed Mater Res 2004;69B:166-172
7. Luthardt RG, Holzhüter M, Sandkuhl O, et al: Reliability and properties of ground Y-TZP-zirconia ceramics. J Dent Res 2002;81:487-491
8. Luthardt RG, Holzhüter M, Rudolph H, et al: CAD/CAM-machining effects on Y-TZP zirconia. Dent Mater 2004;20:655-662
9. Dittmer MP, Kohorst P, Borchers L, et al: Finite element analysis of a four-unit all-ceramic fixed partial denture. Acta Biomater 2008;5:1349-1355
10. Wiederhorn SM: Moisture assisted crack growth in ceramics. Int J Fracture Mech 1968;4:171-177
11. Chevalier J, Olagnon C, Fantozzi G: Subcritical crack propagation in 3Y-TZP ceramics: static and cyclic fatigue. J Am Ceram Soc 1999;82:3129-3138
12. Guo X: Property degradation of tetragonal zirconia induced by low-temperature defect reaction with water molecules. Chem Mater 2004;16:3988-3994
13. Chevalier J, Cales B, Drouin JM: Low-temperature aging of Y-TZP ceramics. J Am Ceram Soc 1999;82:2150-2154
14. Garvie RC, Urbani C, Kennedy DR, et al: Biocompatibility of magnesia-partially stabilized zirconia (Mg-PSZ). J Mater Sci 1984;19:3224-3228
15. Thompson I, Rawlings RD: Mechanical behaviour of zirconia and zirconia-toughened alumina in a simulated body environment. Biomaterials 1990;11:505-508
16. Rosentritt M, Behr M, Gebhard R, et al: Influence of stress simulation parameters on the fracture strength of all-ceramic fixed-partial dentures. Dent Mater 2006;22:176-182
17. Ritter JE, Vigedomine M, Breder K, et al: Dynamic fatigue of indented soda-lime glass as a function of temperature. J Mater Sci 1985;28:2868-2872
18. Addison O, Fleming GJ, Marquis PM: The effect of thermocycling on the strength of porcelain laminate veneer (PLV) materials. Dent Mater 2003;19:291-297
19. Scherrer SS, De Rijk WG: The fracture resistance of all-ceramic crowns on supporting structures with different elastic moduli. Int J Prosthodont 1993;6:462-467
20. Stiesch-Scholz M, Schulz K, Borchers L: In vitro fracture resistance of four-unit fiber-reinforced composite fixed partial dentures. Dent Mater 2006;22:374-381
21. Parfitt GJ: Measurement of the physiological mobility of individual teeth in an axial direction. J Dent Res 1960;39:608-618
22. Heners M: Elektronische Untersuchung zur Reproduzierbarkeit des okklusalen Traumas [Electronic studies on the reproducibility of occlusal trauma]. Dtsch Zahnärztl Z 1977;32: 433-439
23. 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. J Oral Rehabil 2005;32:180-187
24. Marx R, Jungwirth F, Walter PO: Threshold intensity factors as lower boundaries for crack propagation in ceramics. Biomed Eng OnLine 2004;3/41:1-9
25. Deng Y, Lawn BR, Lloyd IK: Characterization of damage modes in dental ceramic bilayer structures. J Biomed Mater Res 2002;63:137-145
26. Fischer H, Weinzierl P, Weber M, et al: Bearbeitungsinduzierte Schädigung von Dentalkeramik [Processing related damages of dental ceramic]. Dtsch Zahnärztl Z 1999;54:484-488
27. Zhang Y, Pajares A, Lawn BR: Fatigue and damage tolerance of Y-TZP ceramics in layered biomechanical systems. J Biomed Mater Res 2004;71B:166-171
28. Kao HC, Ho FY, Yang CC, et al: Surface machining of fine-grain Y-TZP. J Eur Ceram Soc 2000;20:2447-2455
29. Zhang Y, Lawn BR, Rekow ED, et al: Effect of sandblasting on the long-term performance of dental ceramics. J Biomed Mater Res 2004;71B:381-386

30. Scherrer SS, Denry IL, Wiskott HWA: Comparison of three fracture toughness testing techniques using a dental glass and a dental ceramic. Dent Mater 1998;14:246-255
31. Kosmac T, Oblak C, Jevnikar P, et al: The effect of surface grinding and sandblasting on flexural strength and reliability of Y-TZP zirconia ceramic. Dent Mater 1999;15: 426-433
32. Chevalier J, Olagnon C, Fantozzi G, et al: Crack propagation behavior of Y-TZP ceramics. J Am Ceram Soc 1995;78:1889-1894
33. Sindel J, Petschelt A, Grellner F, et al: Evaluation of subsurface damage in CAD/CAM machined dental ceramics. J Mater Sci-Mater Med 1998;9:291-295
34. Kosmac T, Oblak C, Jevnikar P, et al: Strength and reliability of surface treated Y-TZP dental ceramics. J Biomed Mater Res 2000;53:304-313
35. Grant KL, Rawlings RD, Sweeney R: Effect of HIPping, stress and surface finish on the environmental degradation of Y-TZP ceramics. J Mater Sci Mater Med 2001;12:557-564
36. Huang H: Machining characteristics and surface integrity of yttria stabilized tetragonal zirconia in high speed deep grinding. Mater Sci Eng A Struct 2003;345:155-163
37. Kondoh J: Origin of the hump on the left shoulder of the X-ray diffraction peaks observed in Y_2O_3-fully and partially stabilized ZrO_2. J Alloy Compound 2004;375:270-282
38. Deville S, Chevalier J, Gremillard L: Influence of surface finish and residual stresses on the aging sensitivity of biomedical grade zirconia. Biomaterials 2006;27:2186-2192
39. Koh Y-H, Kong Y-M, Kim S, et al: Improved low-temperature environmental degradation of yttria-stabilized tetragonal zirconia polycrystals by surface encapsulation. J Am Ceram Soc 1999;82:1456-1458
40. Deville S, Chevalier J, Attaoui HE: Atomic force microscopy study and qualitative analysis of martensite relief in zirconia. J Am Ceram Soc 2005;88:1261-1267
41. Guazzato M, Quach L, Albakry M, et al: Influence of surface and heat treatments on the flexural strength of Y-TZP dental ceramic. J Dent 2005;33:9-18
42. Beuer F, Steff B, Naumann M, et al: Load-bearing capacity of all-ceramic three-unit fixed partial dentures with different computer-aided design (CAD)/computer-aided manufacturing (CAM) fabricated framework materials. Eur J Oral Sci 2008;116:381-386
43. Lüthy H, Filser F, Loeffel O, et al: Strength and reliability of four-unit all-ceramic posterior bridges. Dent Mater 2005;21:930-937
44. Curtis AR, Wright AJ, Fleming GJ: The influence of simulated masticatory loading regimes on the bi-axial flexure strength and reliability of a Y-TZP dental ceramic. J Dent 2006;34:317-325
45. Curtis AR, Wright AJ, Fleming GJ: The influence of surface modification techniques on the performance of a Y-TZP dental ceramic. J Dent 2006;34:195-206