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The effect of mechanical fatigue and accelerated ageing on fracture resistance of glazed monolithic zirconia dental bridges

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ABSTRACT

The fracture resistance of glazed four-unit posterior dental bridges after mechanical fatigue testing and artificial ageing was evaluated. Thirty identical monolithic bridges were fabricated from "translucent" zirconia and divided into three groups of ten. The first group was monotonically loaded to fracture; the second group was dynamically loaded in water (0–300 N, 10^6 cycles) prior to fracture, while the third group was first subjected to accelerated ageing (in a diluted acetic acid solution at 134 °C for 12 h), then to mechanical fatigue and finally monotonically loaded to fracture. Two tested bridges did not survive 10^6 cycles of dynamic loading, one in Group 2 and one in Group 3. Mean monotonic fracture loads (N) were: Group 1: 547.3 ± 66.3, Group 2 (n = 9): 465.2 ± 118.0, and Group 3 (n = 9): 408.8 ± 58.9. According to the fractographic analysis the glaze embrittlement during artificial ageing and the stress corrosion during fatigue loading are proposed to be responsible for the reduced fracture resistance.

1. Introduction

In restorative dentistry, the replacement of metal-based FDPs with all-ceramic crowns and bridges has been driven by the improved aesthetics and excellent tissue compatibility achieved using tooth-colored, metal-free systems [1]. Amongst load-bearing dental ceramics, yttria partially stabilized tetragonal zirconia (3Y-TZP) ceramics hold a unique place: at room temperature the tetragonal grains in sintered ceramics are metastable and can be transformed by stress into the stable monoclinic form thereby developing transformation toughening resulting in superior strength and fracture toughness. On the other hand, the metatastability of tetragonal grains makes 3Y-TZP ceramics susceptible to hydrothermally induced transformation (known as LTD or ageing). When exposed to moist environments at slightly elevated temperatures over long periods, the surface of Y-TZP transforms spontaneously from the tetragonal (t) to monoclinic (m) structure [2]. Due to a considerable volume expansion associated with the t-m transformation, this process is accompanied by extensive micro-cracking, which may lead to severe strength degradation or spontaneous fracture which, in turn, may deteriorate the mechanical properties [2–4]. Unfortunately, conventional 3Y-TZP ceramics are opaque and unless they are colored, they have a white appearance. Being least translucent of all contemporary dental ceramic materials for all-ceramic dental restorations this type of dental zirconia is mainly used for the frameworks of fixed dental prostheses that are veneered with porcelain to achieve the final shape also providing matching hardness and elastic modulus with those of the hard tooth tissue [5]. Since in this way the translucency of natural teeth cannot be mimicked in every detail, the clinical use of dental zirconia (3Y-TZP) ceramics has been limited to multi-unit posterior bridges, implant supported crowns and other fixed dental prostheses (FDPs) in the less visible side areas, where rather than aesthetics the mechanical properties are paramount. Conversely, in the frontal area, *e.g.* in prosthetic rehabilitation of upper incisors, where dental display is of utmost importance, more translucent materials such as leucite or lithium-disilicate glass ceramic or glass-infiltrated ceramics are being preferred.

The clinical use of porcelain veneered zirconia all-ceramic FDPs is often associated with chipping or fracturing of the veneering porcelain. Chipping of porcelain veneered to ceramics occurs much more frequently than chipping of porcelain bonded to metal-ceramic FDPs [6]. This is due to higher chemical inertness, less favourable wetting ability and considerably lower thermal conductivity of ceramics, as compared to metals [7]. Minor chipping damages can be temporarily repaired, but in most cases the whole

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prosthetic work has to be replaced. Less porcelain chipping and delamination can be attained with uniformly thick layer of veneering porcelain fired onto anatomically shaped zirconia framework provided that the thermal expansion coefficient of the selected veneering porcelain is s adapted to the zirconia ceramic [8].

Frequent necessity for repair or replacement of zirconia-based FDPs due to porcelain delamination has ultimately led to the development of full-contour monolithic zirconia restorations without veneering porcelain. The advantages of monolithic zirconia restorations over glass-ceramic and porcelain veneered zirconia restorations are higher strength, reduced tooth preparation and faster completely computer-aided manufacturing [9,10].

The introduction of monolithic (full-contour) dental zirconia restorations in contemporary prosthodontics has only been possible after the development of ceramics with improved translucency. Compared to conventional 3Y-TZP, this type of zirconia ceramic exhibits finer grains, reduced porosity and a smaller amount of alumina commonly added to conventional ceramics to suppress LTD. With the improved translucency of dental zirconia their clinical indications expanded. Besides FDPs in high loaded posterior area, monolithic Y-TZP ceramics is currently used also in aesthetically demanding areas (including prosthetic rehabilitation of upper incisors). Recently, complex implant-prosthetic restorations have also been successfully completed in monolithic zirconia [11].

Zirconia dental restorations, that are veneered with a feldspatic porcelain, are commonly glazed in the final stage of the manufacturing process to become smooth and less abrasive. The aim of glazing is to seal the open pores in the surface of a fired porcelain. Dental glazes are composed of colorless glass powder, applied to the fired porcelain or glass ceramic to produce a glossy surface [12]. There is a lack of agreement in the literature on the necessity of glazing of monolithic Y-TZP ceramics. On the one side the glaze may reduce wear of the antagonist teeth and prevent low temperature degradation by protecting the material from direct contact with humid oral environment. On the other side, the thin layer of the soft glaze is likely to be abraded and/or peeled away during mastication. For this reason the opponents do not believe that glazing is effective and advocate for fine polishing after occlusal adjustments [13–15].

Zirconia based all-ceramic FPDs are expected to be in service for at least 7–10 years under clinical conditions, where they are exposed to cyclic mechanical and thermal loadings in the chemically aggressive environment of the oral cavity. Consequently their strength tends to diminish steadily with time, from stress corrosion and fatigue as well as other mechanisms, *e.g.* during mechanical surface treatment and/or enhanced low temperature degradation (LTD), pertaining toY-TZP ceramics. According to an extensive *in-vitro* study on the combined effect of mechanical surface treatment and accelerated ageing on the survival rate of conventional 3Y-TZP ceramic upon fatigue testing in an artificial saliva solution the ceramic indeed undergoes strength degradation upon surface damage and cyclic fatigue loading. The extent of the strength degradation was greatest for surface damaged (ground) and artificially aged specimens, while LTD alone had a minimal effect on the strength and fatigue reliability [16].

Clinically reported fractures originating in the conventional zirconia framework are relatively scarce [17,18] and there is no firm evidence of LTD-initiated fracture so far. However, with the introduction of porcelain-free monolithic FDPs in clinical practice, of Y-TZP sintered at higher temperatures, this situation may drastically change, as these FDPs will be directly exposed to the moist environment of the oral cavity. Current *in-vitro* studies on durability of experimental zirconia-based FDPs were performed with porcelain-veneered bridges [19–22], but no literature is available on testing of monolithic long span dental bridges made from yttria stabilized zirconia ceramic combined with accelerated ageing.

Therefore the purpose of this study was to evaluate the effect of mechanical fatigue with and without accelerated ageing on fracture resistance and reliability of glazed posterior monolithic zirconia four-unit dental bridges. This *in-vitro* study will also elucidate the impact of surface glaze on the performance of tranlucent zirconia material by combinining *in-vitro* laboratory conditions with basic requirements of the clinical situations.

2. Experimental procedures

2.1. *Manufacturing of models for* in-vitro *fatigue testing*

The experimental four-unit dental bridges were produced *in vitro* on an artificial dental model (Frasaco, Tettnang, Germany). The two abutment teeth (first lower premolar and second lower molar) were ground for 1.0–1.5 mm in depth, according to standardized dental preparation procedures.

Silicon impressions of the prepared teeth were taken and the molds were produced. Teeth analogues were cast from polyurethane resin (PUR; Alpha Schütz Dental, Germany). The roots of the teeth were coated with 1 mm thick silicon layer (Erkoskin, Erkodent, Germany) to simulate periodontal mobility. In order to achieve a constant silicon layer, the roots were dipped in a wax bath before the PUR base model was manufactured.

2.2. *Manufacturing of monolithic zirconia bridges*

The master model was optically scanned (inEos X5, Sirona, Germany) and the monolithic zirconia ceramic bridge was computer-aided designed. A total of 30 identical four-unit bridges with at 9 mm^2 cross-sections of the connectors, were milled from translucent zirconia blanks (inCoris TZI, Sirona, Germany) in anatomic design using a commercial milling unit (Sirona In-Lab MC X5, Germany) equipped with diamond rotary tools. The bridges were sintered in an electric furnace (in Fire, HTC, Germany) at a temperature of 1510 °C for 2 h. A commercial glaze (Vita Akzent, Germany) was applied with a brush technique to the outer surface. Bridges were glazed according to the manufacturer's instructions and fired for 2 min at 850 °C. After glazing, the bridges were randomly divided into three groups of 10.

2.3. *Mechanical fatigue and fracture load testing*

Prior to mechanical testing the bridges were cemented with a resin adhesive (Panavia 21 TC, Kurraray, Japan) onto the supporting fixture as part of a specially designed experimental set-up mimicking the real clinical situation (Fig. 1). The supporting fixture was made of PUR with elastic properties similar to bone and dentin.

All mechanical tests were performed in deionized water at 37 °C using a servo-hydraulic testing system (Instron 8871, Instron, High Wycombe, UK). The load was vertically transferred onto the middle pontic center *via* a stainless steel ball (diameter 6 mm) on an interposed macrolon foil (Fig. 2). The control group (Group 1) was monotonically loaded to fracture at a crosshead-speed of 1 mm/min and the fracture load was registered. Group 2 was subjected to mechanical sinusoidal cyclic loading ranging from 0 to 300 N at a frequency of 15 Hz. A fatigue cycle limit of 10^6 was set and the specimens that survived the mechanical fatigue testing were subsequently monotonically loaded to fracture. The third group (Group 3) was first subjected to accelerated ageing in 4% acetic acid solution (adopted for testing the chemical solubility of dental ceramics, as specified in ISO 6872:1996(E)) at 134 °C for 12 h. After ageing the bridges were mechanically cycled under same conditions as in Group 2 and finally monotonically loaded to fracture.

After mechanical testing, the fracture surfaces of the broken bridges were first inspected by optical lenses (Stereo Discovery V8,