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Original article

Fatigue performance and failure load of minimally-invasive occlusal veneers made of lithium disilicate and composition-gradient multilayered zirconia: An *in vitro* study

Lea S. Prott [a,b,*], Monique Harlaß [a], Alexander Marksteiner [a], Frank A. Spitznagel [a], Robert Langner [c,d], Yu Zhang [b], Markus B. Blatz [b], Petra C. Gierthmuehlen [a]

[a] Department of Prosthodontics, Medical Faculty and University Hospital Düsseldorf, Heinrich-Heine-University, Düsseldorf, Germany, [b] Department of Preventive and Restorative Sciences, Penn Dental Medicine, University of Pennsylvania, Philadelphia, USA, [c] Institute of Systems Neuroscience, Medical Faculty, Heinrich-Heine-University Düsseldorf, Düsseldorf, Germany, [d] Institute of Neuroscience and Medicine, Brain and Behavior (INM-7), Research Center Jülich, Jülich, Germany

Abstract

Purpose: To investigate the effect of ceramic material (lithium disilicate, LDS vs. composition-gradient multilayered zirconia [4Y-PSZ and 5-PSZ], Z) and ceramic layer thickness (0.5 mm, 1.0 mm, and 1.5 mm) on fatigue performance and failure load of occlusal veneers on molars.

Methods: Seventy-two CAD-CAM-fabricated occlusal veneer restorations (IPS e.max CAD; IPS e.max ZirCAD Prime Esthetic, Ivoclar Vivadent) were divided into six groups (n=12, LDS-1.5, LDS-1.0, LDS-0.5; Z-1.5, Z-1.0, Z-0.5). Restorations were adhesively cemented (Variolink Esthetic DC, Ivoclar Vivadent) to dentin-analogue composite dies (Z100, 3M ESPE) and exposed to thermomechanical fatigue (1.2 million cycles, 49 N, 1.6 Hz, 5-55° C). Single-load-to-failure was tested with a universal testing machine. Data were analyzed using ANOVA with Tukey post-hoc tests and *t*-tests ($P < 0.05$).

Results: The overall success rate across all materials and layer thicknesses was 91.7%. Half of the specimens of group Z-0.5 revealed cracks after chewing simulation. Occlusal veneers fabricated from LDS withstood significantly higher failure loads than gradient multilayered zirconia veneers in all tested thicknesses. The mean failure load values led to the following ranking: 3194 N (LDS-0.5)>2683 N (LDS-1.0)>2338 N (LDS-1.5)>1744 N (Z-1.5)>1310 N (Z-0.5)>1198 N (Z-1.0).

Conclusions: Ultrathin LDS occlusal veneers outperformed thin and standard thick counterparts, as well as gradient multilayered zirconia veneers at all thickness levels. Ultrathin gradient multilayered zirconia occlusal veneers were prone to cracks during thermomechanical fatigue. Individual mechanical properties need to be considered when aligning the restoration within the multilayered zirconia blank.

Keywords: Ceramics, Occlusal veneer, Ceramic thickness, Fatigue, Computer-aided design

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1. Introduction

During the last decades, tooth wear has grown into an oral health problem with pathological progression and an increasing global prevalence ranging between 26.9% and 90% in permanent teeth[1,2]. In modern society, pathological tooth wear has a multifactorial etiology[3,4], often related to the presence of bruxism combined with chemical processes caused by lifestyle changes such as acid-related food, drink habits and gastro-esophageal reflux[4–6]. Since younger patients are mainly affected, extensive preparations resulting in further loss of sound tooth substance should be avoided. With continuous advancements in adhesive technology and computer-aided design and computer-aided manufacturing (CAD-CAM) materials, minimally invasive concepts with non-retentive restoration designs were introduced[7]. Although several *in vitro* studies have confirmed that the thickness of occlusal veneers composed of lithium disilicate (LDS) can be successfully reduced to 0.7 mm and

WHAT IS ALREADY KNOWN ABOUT THE TOPIC?

» *Minimally invasive occlusal veneers are viable alternatives to conventional full-coverage restorations. While 4Y-PSZ zirconia shows promising results, failures have been reported in 5Y-PSZ veneers. No study has investigated the behavior of occlusal veneers fabricated from the transition layer of multilayered zirconia (4Y- and 5Y-PSZ).*

WHAT THIS STUDY ADDS?

» *Lithium disilicate occlusal veneers exhibited significantly higher failure loads than multilayered zirconia veneers fabricated from the transition layer in all thickness levels. Ultrathin (0.5 mm) multilayered zirconia veneers were prone to crack formation under fatigue, emphasizing the importance of material selection based on material properties.*

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***Corresponding author:** Lea Sophia Prott, Department of Prosthodontics, Medical Faculty and University Hospital Düsseldorf, Heinrich-Heine-University, Moorenstraße 5, 40225 Düsseldorf, Germany.
E-mail address: leasophia.prott@med.uni-duesseldorf.de

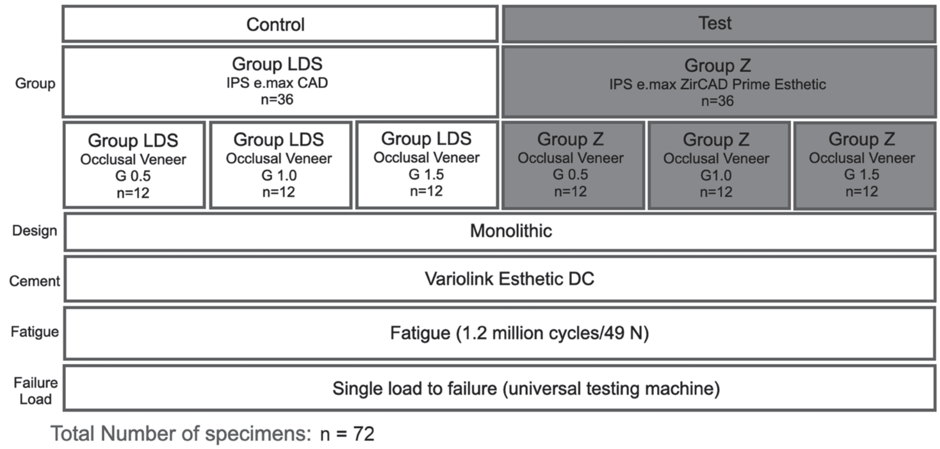

Fig. 1. Test-set up. LDS: lithium disilicate; Z: composition-gradient multilayered zirconia.

less[8–10], manufacturers do not recommend reducing the thickness to less than 1 mm[11].

Due to its beneficial esthetic and mechanical properties, LDS is a widely used ceramic material for the fabrication of occlusal veneers[12]. Its glassy matrix facilitates excellent bondability, while its flexural strength (400-600 MPa) is moderate compared to traditional dental zirconia (3Y-TZP; 3 mol% yttria stabilized zirconia polycrystals; 800-1200MPa)[13]. With improvements in the translucency of dental zirconia, an increased yttria content (4Y-PSZ and 5-PSZ, respectively, 4 mol% and 5 mol% yttria partially stabilized zirconia) resulted in a higher amount of cubic phase particles and, thus, in reduced flexural strength (between 500-900 MPa)[13–15]. These mechanical and esthetic modifications made the translucency and strength of ultra-translucent zirconia more akin to that of LDS and these ceramics more suitable for the fabrication of occlusal veneers[16,17]. Despite material advancements, the use of zirconia for occlusal veneers is currently not endorsed by manufacturers and remains outside the scope of approved clinical indications[18].

Recently, multilayered zirconia discs with a strength and color gradient merging the properties of different zirconia compositions were introduced to the dental market[19,20]. Especially in the posterior area where higher masticatory forces occur, these composition-gradient multilayered zirconia ceramics may be a preferable choice compared to LDS. Restoration success, however, depends on several factors, such as preparation design, occlusal forces, type of cement, restoration thickness, mechanical properties, and applied bonding techniques[7]. It is well known that conventional etching-silane treatment is not effective for the bonding of zirconia. Instead, air-particle abrasion is recommended to promote mechanical retention and decontaminate the bonding surface[21,22]. A previous finite element analysis reported that when bonded to enamel, the load bearing capacity of minimally invasive LDS onlays (0.6-1.4 mm) can exceed 70% of that of zirconia[23]. In the challenging condition of an ultrathin occlusal veneer when the bonding area is low, the preparation design non-retentive, and the thickness reduced by up to 0.5 mm, LDS may benefit from its excellent bondability and occlusal veneers composed of LDS may withstand higher failure loads than their multilayered zirconia counterparts.

Studies comparing the failure load of occlusal veneers made of gradient multilayered zirconia are limited[24]. No research has yet evaluated the use of multilayered zirconia composed of 4Y-PSZ and 5-PSZ for the fabrication of occlusal veneers. Furthermore, little is known about the transition layer. Whether it exhibits a strength gradient between 4Y-PSZ and 5-PSZ remains elusive, since it is technologically challenging to experimentally determine strength gradients in the transition zone. Consequently, further investigation is needed to assess the behavior of these restorations, particularly when fabricated with reduced layer thickness from the transition layer.

This *in vitro* study aimed to investigate the mechanical behavior of occlusal veneers fabricated from gradient multilayered zirconia (4Y-PSZ and 5-PSZ) and LDS in varying thicknesses (0.5 mm, 1.0 mm, and 1.5 mm). The tested research hypotheses were that (i) ceramic material and (ii) ceramic layer thickness would affect fatigue performance and the failure load of posterior occlusal veneer restorations.

2. Materials and Methods

A total of 72 specimens were divided into two groups (Lithium disilicate, LDS: IPS e.max CAD, Ivoclar Vivadent, Schaan, Liechtenstein; Composition-gradient multilayered zirconia, Z: IPS e.max ZirCAD Prime Esthetic, Ivoclar Vivadent) according to the type of ceramic material. Both groups were further divided into subgroups (n = 12) according to their ceramic layer thickness (0.5 mm, 1.0 mm, and 1.5 mm) (**Fig. 1**).

2.1. Specimen preparation

A maxillary first molar of a typodont model (Frasaco-model, Frasaco, Tettnang, Germany) was prepared to obtain three master dies with an occlusal reduction of 0.5 mm, 1.0 mm, and 1.5 mm. The preparation design was non-retentive (occlusal reduction exclusively) with two additional diagonally set shallow notches (0.2 mm depth) that were prepared to facilitate restoration positioning during adhesive cementation (**Fig. 2**). The preparation was performed by an experienced prosthodontist using diamond burs (no. 8370314 035, no. 370314 035, no. 8801314 023, Komet, Lemgo, Germany) under 4.5-fold magnification and with high-speed instrumentation under air-water-spray cooling. Before preparation, silicone impressions (TwinDuo, Picodent, Wipperfürth, Germany) were taken and

Fig. 2. Non-retentive preparation design of occlusal veneer including an occlusal reduction and two additional diagonally set shallow notches.

sectioned in a bucco-palatal direction to control the exact tooth substance removal. A periodontal probe was used to verify the preparation depth.

To fabricate 72 dentin-analogue abutments, impressions of the three master dies were taken using a polyvinylsiloxane material (Identium, Kettenbach, Eschenburg, Germany). The resulting negative molds were then filled with 1.5 mm-thick layers of a resin-based composite (Filtek Z100, 3M ESPE, Neuss, Germany), which has an elastic modulus of 18 GPa – comparable to that of human dentin (16-18 GPa)[25,26]. Each layer was light cured for 20 s (Bluephase G4 with 1200 mW/cm^2, Ivoclar Vivadent). Resin dies were immersed in distilled water and stored in an incubator (Universalschrank UF55, Memmert, Schwabach, Germany) at 37° C for 3 to 5 weeks. Afterwards, all resin dies were placed into a self-curing epoxy resin (RenCast CW20/ Ren HY 49, Huntsman Advanced Materials, TX, USA).

2.2. *Fabrication of occlusal veneer restorations*

After the scanning of typodont models including the three different master dies (PrograScan PS5 v3.2, Ivoclar Vivadent), occlusal veneer restorations were designed with respective thicknesses (Dental CAD 3.1 Rijeka, Exocad GmbH) (**Fig. 3**). ZirCAD Prime Esthetic blanks consist of two different raw materials: the dentin part made of 4Y-PSZ and the incisal part made of 5Y-PSZ. The incisal zone is 3 mm, followed by a seamless transition zone of 4 mm (no discrete layers), while the rest of the disc height is formed by the dentin zone[18]. All restorations were positioned in the transition layer of the gradient multilayered zirconia blank, 3.3 mm from the upper border of the blank to the top surface of the veneer (**Fig. 4**). Restorations were finally evaluated in a CAM software (PrograMill CAM Software v5.2.008.00, Ivoclar Vivadent) and milled in a five-axis milling machine (PrograMill PM7 v98.130, Ivoclar Vivadent) out of either partially crystalized LDS CAD/CAM blocks (IPS e.max CAD LT, Ivoclar Vivadent) or biscuit fired translucent multilayered gradient zirconia blanks (IPS e.max ZirCAD Prime Esthetic, Ivoclar Vivadent). A single master dental technician fabricated all restorations in accordance with the manufacturer's guidelines. During the fabrication process, layer thickness was carefully controlled with a caliper (Kroeplin GmbH, Schlüchtern, Germany).

2.3. *Adhesive cementation*

Following the manufacturer's instructions, the intaglio surfaces of all LDS restorations were etched with 4.9% hydrofluoric (HF) acid (IPS Ceramic Etching Gel, Ivoclar Vivadent) for 20 s, thoroughly air-water sprayed, and dried with an oil-free air stream. A silane coupling agent (Monobond Plus, Ivoclar Vivadent) was applied and dried gently for 60 s. The intaglio surface of all gradient multilayered zirconia restorations was air-particle abraded with 50 µm aluminum oxide at a pressure of 1 bar for 10 s, and a special ceramic primer containing functional phosphate monomers (Monobond Plus, Ivoclar Vivadent) was applied for 60s. Resin dies were cleaned using pumice powder (Picodent, Wipperfürth, Germany). Then the surface was rinsed with air-water spray, dried using oil-free air, and treated with 70% ethanol. A light-curing dental adhesive (Adhese Universal, Ivoclar Vivadent) was applied for 20 s. Afterwards, resin dies were air-dried again and light-cured for 20 s (Bluephase G4 with 1200 mW/cm^2, Ivoclar Vivadent). For adhesive cementation, a dual-curing resin-based composite cement (Variolink Esthetic DC, Ivoclar Vivadent) was applied to the intaglio surface of the restoration. A secure positioning was ensured by the diagonal set of occlusal notches. After the removal of excess cement, restoration margins were covered with glycerin-gel (Liquid Strip, Ivoclar Vivadent) and light-curing was performed for 20 s from each side (Bluephase G4, Ivoclar Vivadent). To achieve a complete curing[27], specimens were subsequently stored in distilled water for 24 h at 37° C in an incubator (Universalschrank UF55, Memmert).

2.4. *Fatigue analysis*

All specimens were exposed to cyclic mechanical loading with simultaneous thermocycling (5-55° C, dwell time 120 s) in a dual-axis chewing simulator (CS 4.8 professional line, SD Mechatronik, Feldkirchen-Westerham, Germany) with a load of 49 N at 1.6 Hz for 1.2 million cycles. This *in vitro* test set-up mimicked an aging process equivalent to 5 years of clinical use under artificial conditions and has been validated as a reliable method for assessing fatigue[28–30]. Steatite spheres (r = 3 mm, Hoechst CeramTec, Wunsiedel, Germany) that moved 0.5 mm laterally downwards from the disto-palatal cusp to the central fissure during fatigue simulation were used as antagonists[31,32]. During thermodynamic loading, specimens were inspected twice a day for cracks, fractures, or debonding. Survival and success rates after fatigue exposure were calculated. Unharmed specimens were rated as 'success,' whereas specimens that experi-

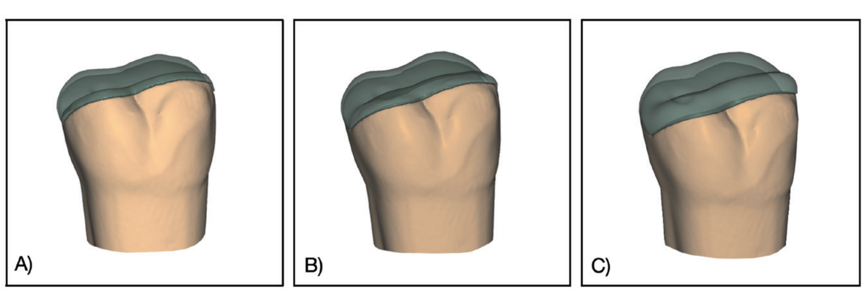

Fig. 3. Restoration design of occlusal veneers in respective thicknesses; A) 0.5 mm, B) 1.0 mm, and C) 1.5 mm.

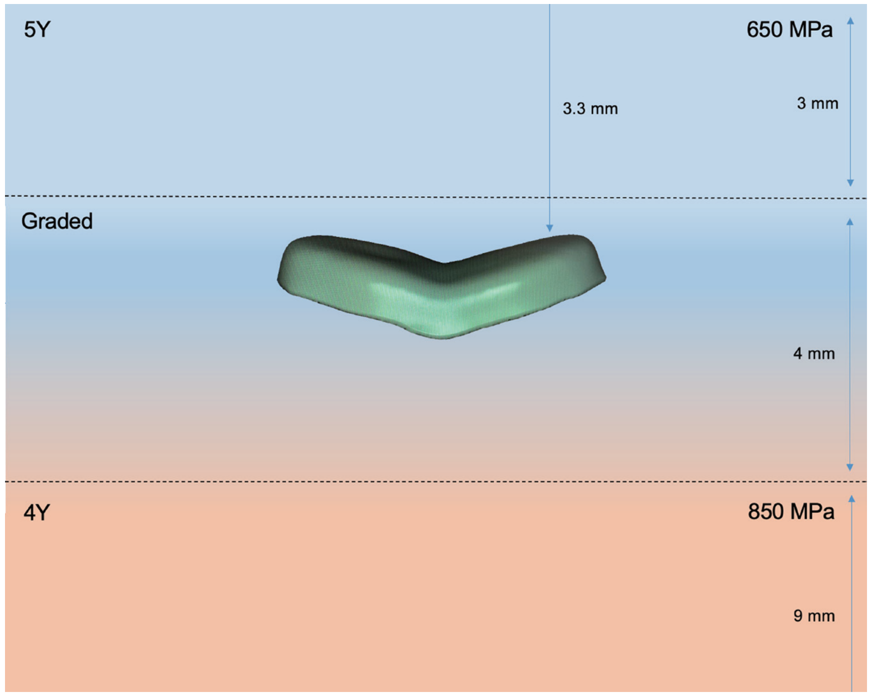

Fig. 4. Veneers of each thickness (0.5 mm, 1.0 mm, and 1.5 mm) were placed within the transition layer of the gradient multilayered zirconia blank, maintaining a distance of 3.3 mm from the upper border of the blank to the top surface of the veneer.

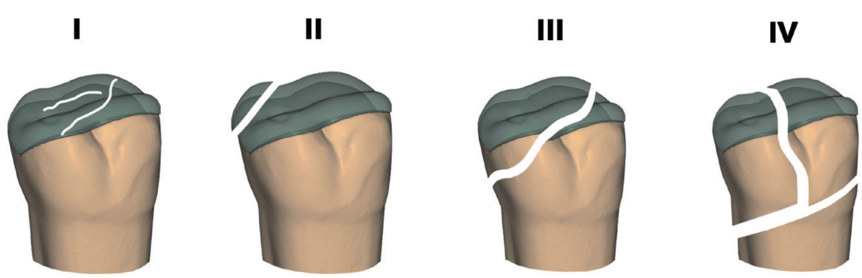

Fig. 5. Failure modes after single load to failure testing: (I) Crack formation within the ceramic. (II) Cohesive fracture within the ceramic, intact tooth. (III) Fracture within ceramic and tooth structure. (IV) Serious/longitudinal tooth fracture involving the root.

enced evident cracks or debonding but were still in function were assessed as 'survival'. Non-survival was when catastrophic fractures occurred[10,33].

2.5. *Single-load-to-failure testing*

After fatigue exposure, all specimens underwent single-cycle load-to-failure (SLF) testing in a universal testing machine (Zwick Z010/TN2S, Zwick Roell, Ulm, Germany). A steel ball indenter (r = 3 mm) was applied axially (crosshead speed of 1.5 mm/min) at the area of the abrasion, i.e., at the same contact point as loaded during thermomechanical fatigue. Specimens were loaded vertically until a fracture occurred. The maximum failure-load was captured through the corresponding test software (TextXpert III, Zwick Roell). In addition, a video camera was used to record the procedure of SLF testing. Failure was considered either as a visible crack or fracture, or a 20% reduction in maximum load (F_{max}) without a visible event.

2.6. *Failure analysis*

Failure analysis was conducted with a polarized light microscope (AxioZoom V.16, Zeiss, Oberkochen, Germany), aiming to identify the failure mode and fracture origin. Failure modes were categorized as follows: (I) Crack formation within the ceramic. (II) Cohesive fracture within the ceramic, intact tooth. (III) Fracture within ceramic and tooth structure. (IV) Serious/longitudinal tooth fracture involving the root (**Fig. 5**)[34].

2.7. *Statistical analysis*

A power calculation using G*Power 3.1.9.7 (Düsseldorf, Germany) was conducted for a 2 x 3 factorial design examining two factors (i) ceramic material (LDS versus gradient multilayered zirconia) and (ii) ceramic layer thickness (0.5 mm, 1.0 mm, 1.5 mm) with respect to statistical testing via ANOVA. The analysis determined that a sample size

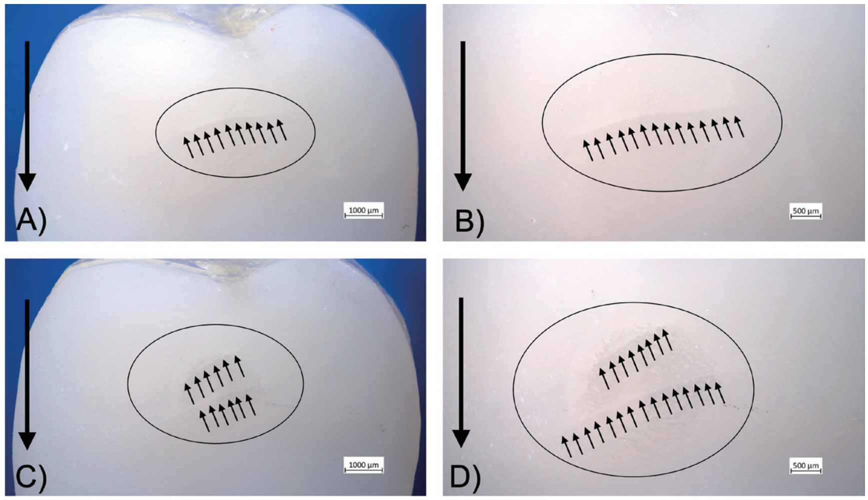

Fig. 6. Light microscope images of 0.5 mm thick gradient multilayered zirconia occlusal veneers (group Z-0.5) showing cracks (small arrows) and wear facets (circle) after fatigue exposure. The bold arrow illustrates the sliding direction during fatigue exposure. A+B): Z-0.5-10 after 381,053 cycles; C+D) Z-0.5-7 after 95,107 cycles.

Table 1. Success and survival rates after simulated fatigue exposure of 5 years

Group	Intact and unharmed specimens after fatigue	Success rate	Overall success rate	Overall survival rate
LDS-1.5	12/12	100%	91.7%	100%
LDS-1.0	12/12	100%		
LDS-0.5	12/12	100%		
Z-1.5	12/12	100%		
Z-1.0	12/12	100%		
Z-0.5	6/12 (six cracks after 2x 95,107 cycles, 381,053 cycles, 630,042 cycles, 656,903 cycles, and 840,513 cycles)	50%		

LDS: lithium disilicate; Z: composition-gradient multilayered zirconia.

of n = 12 per group (n = 72 in total) would provide sufficient power (80%) to detect at least medium-sized effects (Cohen's effect size of f = 0.26) with a two-sided type-I error of $P < 0.05$ for both factors and their interactions.

Data analysis was performed with SPSS 26 (IBM Corp., Armonk, NY, USA). Prior to ANOVA, the Levene test checked for homogeneity of variance. ANOVA was then applied to assess the main effects and interactions of ceramic material and layer thickness. The effect of ceramic layer thickness was further analyzed by separate one-way ANOVAs per material type, followed by post-hoc Tukey tests. The impact of ceramic material was examined separately for each layer thickness level using post hoc two-sample t-tests, preceded by Levene tests to assess the equality of variances. The level of significance was set at $P < 0.05$ (95%-CI) for all tests. Boxplots were generated for descriptive analysis of the data.

3. Results

3.1. Fatigue performance

All restorations showed wear scars after fatigue testing, located in the area between the mesio- and distopalatal cusps as a result of the steatite ball's lateral sliding motion during cyclic mechanical loading. After thermo-mechanical ageing, the tested specimens had an overall survival rate of 100%. No debonding or ceramic fractures occurred during or after dynamic fatigue loading. However, half of the specimens (6/12) of group Z-0.5 (2x after 95,107 cycles, after 381,053 cycles, 630,042 cycles, 656,903 cycles, and 840,513 cycles) showed cracks after chewing simulation (**Fig. 6**), resulting in an overall success rate of 91.7%. Survival and success rates of occlusal veneer restorations are displayed in **Table 1**.

3.2. Single load to failure

The mean failure load values led to the following ranking: 3194 N (LDS-0.5) > 2683 N (LDS-1.0) > 2338 N (LDS-1.5) > 1744 N (Z-1.5) > 1310 N (Z-0.5) > 1198 N (Z-1.0) (**Table 2**). The factor material (LDS vs. Z) showed a significant main effect with respect to failure load [$F(1,66) = 73.9$, $P < 0.001$], whereas ceramic layer thickness (0.5 mm vs. 1.0 mm vs. 1.5 mm) did not [$F(2,66) = 1.4$, $P = 0.247$]. However, the interaction between the two factors (material and layer thickness) was statistically significant [$F(2,66) = 6.2$, $P = 0.004$].

Post-hoc t-tests for separate comparisons between ceramic materials (LDS vs. Z) for each level of ceramic layer thickness revealed

Table 2. Failure load results for all tested groups

Group name	Min	1st Qu	Median	Mean	3rd Qu	Max	SD
LDS-1.5	1785	2068	2245	2338	2607	3500	424
LDS-1.0	1030	2091	2745	2683	3274	4510	931
LDS-0.5	1157	2532	3040	3194	3855	5090	1041
Z-1.5	1450	1521	1617	1744	1967	2669	351
Z-1.0	985	1026	1100	1198	1370	1930	271
Z-0.5	984	1010	1099	1310	1610	2530	472

N: Newton; Min: minimum; 1st Qu: 25th percentile; Median: 50th percentile; 3rd Qu: 75th percentile; Max: maximum; SD: standard deviation, LDS: lithium disilicate; Z: composition-gradient multilayered zirconia.

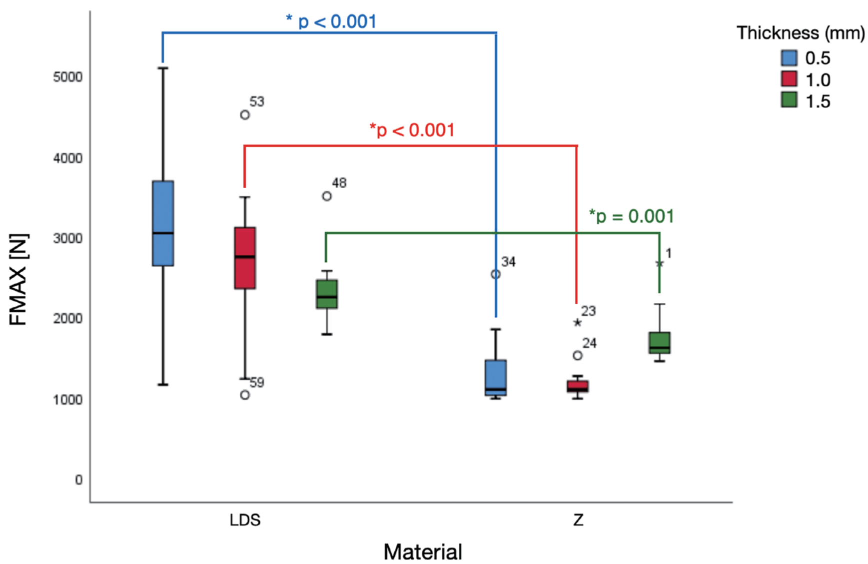

Fig. 7. Boxplot of failure loads (F_{max} in N) shows statistically significant differences between LDS and composition-graded multilayered zirconia occlusal veneers in different layer thicknesses. Statistically significant results (P < 0.05) are marked with an asterisk before the P value. LDS: lithium disilicate; Z: composition-gradient multilayered zirconia.

significant differences on all layer thickness levels (1.5 mm, 1.0 mm, 0.5 mm) in favor of LDS compared to Z (LDS-0.5/Z-0.5: t(22)= -5.7, P < 0.001; LDS-1.0/Z-1.0: t(12.8) = -5.3, P < 0.001; LDS-1.5/Z-1.5: t(22)= -3.7, P = 0.001) (**Fig. 7**). Effects of layer thickness were further analyzed by separate one-way ANOVAs for each ceramic material group (LDS and Z), followed by Tukey's post-hoc tests. In the LDS group, thickness may have an effect [F(2,33) = 3.1, P = 0.057], driven by a significant difference in favor of ultrathin (0.5 mm) veneers compared to their standard thick (1.5 mm) counterparts (P = 0.046), with no other significant differences according to Tukey post-hoc analysis. For Z, thickness also had a significant effect [F(2,33) = 7.1, P = 0.003], but this time, Tukey post-hoc tests revealed significant differences in favor of standard thick (1.5 mm) compared to both thin (1.0 mm) (P = 0.003) and ultrathin (0.5 mm) (P = 0.02) occlusal veneers (**Fig. 8**).

3.3. Failure analysis

Failure analysis after fatigue revealed six specimens of group Z-0.5 with radial cracks beginning from the intaglio cementation surface beneath the contact area. However, they did not propagate to the occlusal contact surface. After SLF, failure modes III (fracture within ceramic and tooth structure) and IV (serious/longitudinal tooth fracture involving the root) were identified as dominant failure modes (**Fig. 9**), with mode IV exclusively occurring in group LDS-0.5. In addition to mode III, specimens of group LDS-1.0 were susceptible to failure mode II (cohesive fracture within the ceramic, intact tooth). Only one specimen of group LDS-0.5 and one of group Z-0.5 fractured in mode I (crack formation within the ceramic) (**Table 3**).

4. Discussion

The present *in vitro* study aimed to analyze the effect of ceramic material and ceramic layer thickness on fatigue performance and failure load of posterior occlusal veneer restorations. The tested research hypotheses were not rejected, as ceramic material and ceramic layer thickness had a statistically significant influence on failure loads. All tested specimens withstood failure loads above physiological chewing forces that clinically range between 289-700 N in the posterior dentition and, thus, revealed higher loading forces than generally required[35]. Nevertheless, half of the specimens of group Z-0.5 did not withstand fatigue exposure unscathed.

Failure analysis identified all occurring cracks as radial cracks developing at the internal cementation surface beneath the contact point without reaching the occlusal surface. The limited bonding quality of zirconia compared to LDS may be of major relevance in the

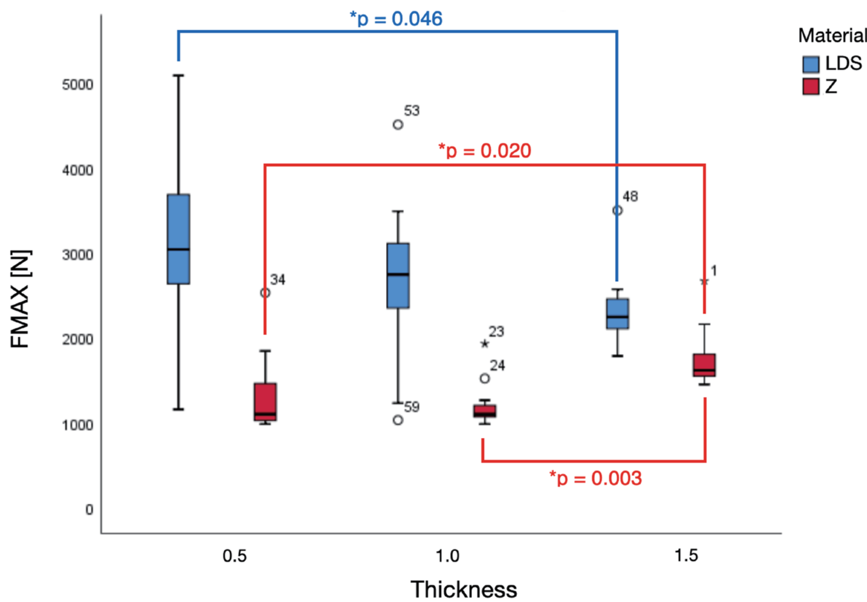

Fig. 8. Boxplot of failure loads (F_{max} in N) shows statistically significant differences between different thicknesses within the test groups (LDS and Z). Statistically significant results ($P < 0.05$) are marked with an asterisk before the P value. LDS: lithium disilicate; Z: composition-gradient multilayered zirconia.

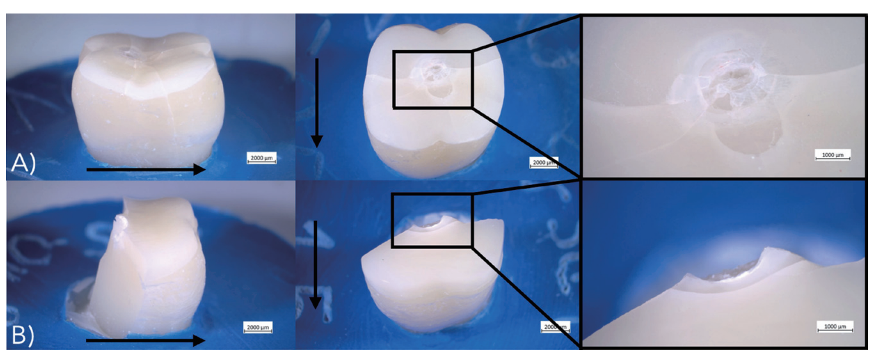

Fig. 9. Light microscope images showing failure modes after single load to failure testing. Ceramic fractures with involvement of the dentin-analogue dies were identified as the dominant failure modes. A) Failure mode III: Fracture within ceramic and tooth structure. B) Failure mode IV: Serious/longitudinal tooth fracture involving the root. The bold arrow illustrates the sliding direction during fatigue exposure.

Table 3. Distribution of failure modes by group (in %). Failure modes were categorized as follows: (I) crack formation within the ceramic, (II) cohesive fracture within the ceramic, intact tooth, (III) fracture within ceramic and tooth structure, (IV) serious/longitudinal tooth fracture involving the root.

Failure mode (%)				
Group name	**I**	**II**	**III**	**IV**
LDS-1.5			100	
LDS-1.0		16.7	83.3	
LDS-0.5	8.3		58.3	33.3
Z-1.5			100	
Z-1.0			100	
Z-0.5	8.3		91.6	

LDS: lithium disilicate; Z: composition-gradient multilayered zirconia.

condition of an ultrathin occlusal veneer compared to a single crown due to the smaller bonding area and the non-retentive preparation design. The critical load for tensile stress-activated flexural radial cracking is predominantly determined by the thickness and flexural strength of the ceramic material[36]. Consequently, radial cracks pose a risk to thin layers with abraded internal surfaces and are reported to be one of the major causes for bulk fractures[37,38]. Air-particle abrasion was used to modify the intaglio zirconia surface, which is sufficient to cause mechanical retention and decontaminate the bonding surface, since it is well known that the conventional etching-silane approach is ineffective for zirconia[21,22]. However, an adjustment of sandblasting parameters is important to avoid excessive microcracks, which may weaken the ceramic material[17]. Therefore, air-particle abrasion with alumina or silica-coated alumina particles (50 µm to 60 µm) at a low pressure (below 2 bar) is recommended[22]. In contrast to zirconia, bonding to LDS ceramics involves HF acid etching and the application of silane coupling agents. HF etching (typically 5% concentration for 20-30 seconds) selectively dissolves the glassy matrix of LDS, creating microporosities and surface grooves that enhance micromechanical retention[39,40]. This is followed by silane application, which chemically bonds the silica-rich surface, forming covalent linkages that facilitate strong adhesion to the resin cement[41]. LDS may derive benefit from this chemical-mechanical approach, which reduces the risk of radial crack propagation in thin restorations by ensuring more uniform stress distribution[42,43].

Similar to fatigue performance, SLF testing further demonstrated a superior performance of LDS across all layer thicknesses (0.5 mm, 1.0 mm, and 1.5 mm). A previous study[13] that used dentin-like composite resin as supporting structure reported a higher load-bearing capacity of LDS (872 N), surpassing 5Y-PSZ (715 N) and even comparable to 4Y-PSZ (864 N). Although ultra-translucent zirconia exhibits significantly lower fracture resistance than traditional conventional dental zirconia, both materials share the same elastic modulus (200 to 210 GPa), which is considerably higher than that of LDS (95 to 105 GPa) and dental hard tissues (enamel: ~70 GPa, dentin: ~18 GPa). It is well known that discrepancies in elastic moduli between ceramics and their supporting structures influence the material's load-bearing capacity[13,23]. This effect may have enhanced the performance of LDS occlusal veneers, as the elastic moduli mismatch is smaller between LDS and dentin relative to zirconia and dentin.

5Y-PSZ still has a low alumina content, around 50-80% of cubic grains, and a remaining part of tetragonal crystals[44]. Compared to Y-TZP, 5Y-PSZ provides an increase in translucency of almost 30% but also a significant drop in flexural strength of 50% with values ranging between 600 to 800 MPa[44–47]. A previous study that compared different layers of multilayered zirconia (IPS e.max ZirCAD MT) reported that the cervical zone (4Y-PSZ) had a significantly better fracture toughness compared to the incisal zone (5Y-PSZ). Another study found that the failure load of 5Y-PSZ crowns differed significantly from 4Y-PSZ and 3Y-TZP, with a wide range of fracture forces between 1211 N (5Y, (thermocycling and mechanical loading (TCML)) and 3952 N (4Y-Multilayer, TCML)[17]. The authors suggested that 5Y-PSZ should be sharply distinguished from other Y-TZP materials and is more comparable to LDS than to "conventional" 3Y-TZP[17]. Compared to single crowns, occlusal veneers have a reduced bonding surface and retention, which may have adversely affected the fatigue and failure load outcomes of the multilayered zirconia specimens. A recently published study investigating the fracture resistance of occlusal veneers made of multilayered zirconia (ZirCAD Prime, 3Y-TZP and 5Y PSZ) confirmed the present findings, with failures (one specimen) and partial failures (7.8% of all specimens) occurring exclusively in veneers within the 5-YTZP layer[24].

In another study, the ceramic material showed no statistically significant effect on the failure loads, which ranged from 2182.53 ± 643.77 to 2505.91 ± 723.01 N for LDS (IPS e.max CAD) and from 1859.88 ± 422.75 to 2285.18 ± 491.06 N for 5Y-PSZ zirconia (Katana UTML, Kuraray Noritake Dental Inc., Japan)[48]. The fatigue protocol differed from the present one, since 120,000 cycles via a 5.4 mm steel piston at a descending speed of 40 mm/s and 1.6 Hz with a previously performed thermocycling (500 cycles, 5-55°C) were applied, making a comparison of both studies difficult[48]. Other *in vitro* studies also reported higher failure loads of 1635 ± 410 N for super high translucent zirconia (Ceramill Zolid FX)[49] and 2382 ± 228 N for high translucent zirconia (Vita YZ HT)[50] occlusal veneers of 0.5 mm thickness. Regarding failure loads of LDS, similar failure loads to the present results have been reported for IPS e.max CAD ultrathin occlusal veneers at 0.3 mm (3591 ± 776 N) and 0.6 mm (2770 ± 598 N) thickness. However, in that study, restorations were bonded to human molars and subjected to mechanical loading with a higher load and higher number of cycles (10 million cycles and 200 N)[9]. Other studies that applied the same thermomechanical fatigue protocol reported lower failure load values of 1178 ± 588 N (0.5 mm) and 1530 ± 440 N (1 mm)[51] as well as 1191 ± 382 N (0.5 mm) and 1851 ± 631 N (1 mm)[50] for occlusal LDS (IPS e.max Press) veneers. Due to differences in preparation design, cementation materials, substrate material, study set-ups, as well as fatigue protocols, direct comparisons to previously published *in vitro* studies are difficult, since these variables can influence failure load values[9].

The high failure load values observed in 0.5 mm LDS veneers can be understood by considering the complex interplay between material properties, veneer thickness, and the biomechanics of the tooth-restoration complex. Firstly, thinner veneers preserve a greater volume of the natural tooth structure, resulting in an enhanced fracture resistance of the tooth restoration complex and a more favorable stress distribution within the tooth structure, compared to extensive preparations that result in greater tooth substance loss[52,53]. Secondly, thinner LDS veneers have been shown to exhibit greater flexural deformation under occlusal loading, thereby augmenting compressive stresses at the contact surface. At the same time, thin veneers transmit the load to the composite substrate, which, in turn, subjects the cementation surface of the veneer to compressive stresses. These compressive stress fields simultaneously suppress the initiation and propagation of cracks from both the contact and flexural surfaces, which is critical for improving the failure load of the restoration[54]. Additionally, the interaction between the LDS veneer and the underlying tooth, in this case, a dentin-analog composite substrate, plays a critical role. LDS has a lower elastic modulus mismatch with natural tooth and composite structures compared to materials like multilayered zirconia. This better elastic compatibility with dentin and enamel or composites reduces stress concentrations at the interface and minimizes the risk of flexural fracture[13].

Limitations of the present study are related to the inherent nature of an *in vitro* study set-up, which cannot fully replicate the complexity of real clinical conditions. The results may have been influenced by the structural elastic response of the material and the specific characteristics of the experimental environment. Typodont teeth and a composite resin were used in place of extracted human molars to ensure standardization; however, they do not perfectly mimic the properties of natural teeth. Pumice powder was used instead of phosphoric acid to clean the surface of the resin. These choices reflect a standardized *in vitro* environment rather than adhesive challenges in clinical practice.

Further preclinical studies are necessary to systematically investigate the failure mechanisms of gradient multilayered zirconia materials with reduced layer thicknesses under highly standardized conditions. Additionally, clinical studies are needed to assess the performance of partial coverage restorations with minimally invasive thicknesses and determine the practical implications of their findings. In clinical conditions, the fracture resistance of thin veneers supported by natural enamel and dentin may be even higher than observed in this *in vitro* study, due to the smaller elastic modulus mismatch between materials. A detailed fracture analysis of specimens that experienced cracks after fatigue and of specimens after SLF exposure needs to be conducted to explain the results of this *in* *vitro* study in greater detail.

5. Conclusions

The ceramic material (lithium disilicate vs. gradient multilayered zirconia) and the ceramic layer thickness significantly affected the failure load of occlusal veneer restorations. Occlusal veneers composed of lithium disilicate withstood significantly higher failure loads than gradient multilayered zirconia veneers in all tested thicknesses (0.5 mm, 1.0 mm, 1.5 mm). Ultrathin (0.5 mm) LDS occlusal veneers significantly outperformed their standard-thick (1.5 mm) counterparts. Gradient multilayered zirconia veneers in 1.5 mm thickness performed significantly better than their thinner (1.0 mm and 0.5 mm) counterparts.

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Conflict of interest disclosure

FS has provided lectures sponsored by Ivoclar Vivadent outside of the submitted work.

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