THE JOURNAL OF PROSTHETIC DENTISTRY

RESEARCH AND EDUCATION

Two-body wear of various microlayering ceramics on zirconia restorations: An in vitro study

Svenja Stuhr, DDS,[a] John Meinen, DT,[b] Michael Reise, Dr rer. nat,[c] Daniel Edelhoff, Prof Dr med dent,[d] Alexis Ioannidis, PD Dr med dent,[e] Katja Gutmair, MSc,[f] and Bogna Stawarczyk, Prof Dr rer biol hum Dipl-Ing(FH)[g]

ABSTRACT

Statement of problem. To prevent chipping of zirconia restorations, thinner veneering techniques like microlayering ceramic (µLC) have been used; however, the wear of these materials requires further investigation.

Purpose. The purpose of this in vitro study was to measure the wear and particle size of µLCs and compare them with veneered and monolithic zirconia and human teeth.

Material and methods. Zirconia crowns (n=12) were microlayered with Ceramotion One Touch 3D (CER), IQ SQIN (IQS), MiYO structure (MIY), and Structure (STR). Veneered zirconia (VEN), monolithic zirconia (MZ), and human teeth (HUM) (n=12) were used as controls. After 60 000, 120 000, and 600 000 masticatory cycles, the 2-body wear of crowns and antagonists (steatite balls) and the particle size of the µLCs was measured. The specimens were qualitatively examined using a light microscope to visualize wear patterns. Data were analyzed using nonparametric statistical methods, including the Spearman and Kendall correlation, Kruskal-Wallis-, Mann-Whitney-U- (α=.05), and ANOVA-type statistics for longitudinal data, with pairwise comparison adjusted by Bonferroni correction (α=.0034).

Results. Material type and number of masticatory cycles affected volumetric loss for crowns and antagonists ($P\leq.003$). MIY showed higher wear than STR ($P\leq.002$), and the remaining µLCs were statistically similar ($P\geq.088$). HUM also fell within the same range of values as CER, STR, IQS, and VEN (P=.191 to.950). HUM showed no wear on antagonists. The wear on MZ could not be quantified. Particle size was determined: MIY<IQS<CER<STR<VEN. A negative correlation was observed between volumetric material loss and particle size ($P<.001$).

Conclusions. The µLCs and VEN showed measurable wear in contrast with MZ; but VEN demonstrated material loss more similar to that of HUM, reflecting physiological behavior. Increased masticatory cycles led to greater wear, risking exposure of the zirconia substrate. (J Prosthet Dent 2026;135:820.e1-e9)

Zirconia has emerged as a popular restorative material in dentistry,[1] primarily because of its excellent mechanical properties and biocompatibility.[2–5] The development of modified zirconias, coupled with advancements aimed at enhancing translucency,[6] has facilitated its application even in esthetically demanding areas using a monolithic approach.[2,7–9] Monolithic zirconia addressed the issue of chipping, which has been reported to occur frequently with veneered zirconia frameworks.[10] As ceramics are sensitive to tensile stresses and while compressive stresses are less critical, applying moderate compressive stress to the veneering part is preferred to reinforce the veneering

Supported by Chemichl, Dentaurum, Estetic Ceram, and GC.
The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.
[a]Graduate student, Department of Prosthetic Dentistry, University Hospital, Munich, Germany.
[b]Dental Technician, Department of Prosthetic Dentistry, University Hospital, Munich, Germany.
[c]Head of Production and Development, Estetic Ceram, Triesen, Liechtenstein.
[d]Professor and Department Head, Department of Prosthetic Dentistry, University Hospital, Munich, Germany.
[e]Head of Fixed Reconstructive Dentistry, Department of Reconstructive Dentistry, University of Zurich, Zurich, Switzerland.
[f]Research Associate, Institute for Medical Information Processing, Biometry and Epidemiology, University Hospital, Munich, Germany.
[g]Professor, Department of Prosthetic Dentistry, University Hospital, Munich, Germany.

Clinical Implications

Microlayering ceramics and veneered zirconia exhibit measurable wear under simulated mastication; however, veneered zirconia demonstrated wear behavior most similar to natural teeth. Material selection and anticipated functional load should therefore be carefully considered to minimize wear.

ceramic and increase fracture resistance.[11] Therefore, thermal properties such as the coefficient of thermal expansion (CTE) should be matched between framework and veneering ceramic to prevent tensile stress and thus reduce failures.[11–13]

Despite the advantages of monolithic zirconia, its minimal wear remains a challenge,[14–16] as it fails to replicate natural tooth wear and limits physiological occlusal adaptation.[17,18] Wear is a multifactorial process characterized by material removal by abrasive food particles, contact between antagonists during mastication, and parafunctional habits.[17,19–21] Factors such as the type of antagonistic material, enamel thickness and hardness, as well as neuromuscular forces significantly influence wear dynamics.[17,19–21] Additionally, mechanical properties including hardness, frictional resistance, and fracture toughness have been reported to play crucial roles in determining wear behavior.[17,19,22,23] The abrasive wear rate of human enamel has been reported to be approximately 30 to 40 µm per year.[17,19,24] Ideally, dental materials should exhibit wear characteristics that closely mimic those of human enamel to match the physiological rate.[25] Microlayering represents a promising alternative to traditional veneering for zirconia restorations.[7] This method involves applying microlayering ceramics (µLC) in minimal thicknesses (0.1 to 0.6 mm) onto anatomically contoured or minimally reduced zirconia frameworks.[7,26] By combining the benefits of monolithic applications, such as reduced removal of tooth structure, with enhanced esthetics akin to veneered crowns, microlayering may also address concerns related to chipping.[7]

Despite its growing popularity, the body of research on the wear behavior of µLC is limited. Furthermore, the particle size of the µLC may correlate with its wear behavior or surface roughness. Previous research has shown that surface roughness impacts wear (of crowns and antagonists).[27] While clinical trials are essential for reflecting the complexities of oral wear, they often entail high costs and time commitments; moreover, certain parameters like mastication forces and environmental factors can be challenging to monitor and control effectively.[17,19] In contrast, laboratory investigations allow for the examination of specific variables under standardized conditions, facilitating comparability across studies.[17] Parameters such as mastication force and frequency, lateral movement, thermal loading, type of abutment, and antagonists can affect the results,[28–30] and crown geometry, abutment material, and antagonist material have been reported to influence 2-body wear[25]; however, the number of masticatory cycles also has an impact on wear.[31,32] In a mastication simulator, as used in laboratory tests, these parameters can be set and monitored. The aim of this study was to investigate the wear behavior of µLCs in relation to their particle size. The null hypothesis was that no significant differences in wear behavior would be found between the investigated groups (microlayered and veneered zirconia crowns, monolithic zirconia, and human teeth).

MATERIAL AND METHODS

In total, 84 crowns and 84 antagonists were analyzed. Four different µLCs (CER, IQS, MIY, STR) were tested and applied on anatomic zirconia frameworks. For control, anatomic zirconia was polished, reduced zirconia frameworks were veneered, and human molars were used (Table 1, Fig. 1). A total of 12 mandibular first zirconia molar crowns per group were milled (CeraMill Motion 2; Amann Girrbach AG) for the µLCs and MZ groups. The sample size of 12 per group was based on previous studies that demonstrated statistically significant differences with a similar sample size.[14,25,33] A master STL file with medium cusp inclination was selected based on a previous study.[25] The buccal wall thicknesses was set to 1.5 mm, the lingual to 1.5 mm, the occlusal to 1.0 mm, and the proximal to 2.0 mm.[25] For the veneered group, a µLC crown was scanned, followed by an occlusal and lateral cutback of 1.2 mm tapering toward the cervical region (Fig. 2). All crowns were sintered to a final temperature of 1500 °C with a holding time of 2 hours according to the manufacturers' instructions.

To prepare the microlayering and veneering areas of the zirconia framework, they were airborne-particle abraded for 10 seconds from 10 mm at 0.05 MPa and at an angle of 45 degrees using a device (Basic quattro IS; Renfert GmbH) with 50-µm alumina powder (Orbis Dental). Following this procedure, the crowns were cleaned in an ultrasonic bath (Sonorex Digitec; Bandelin electronic GmbH & Co KG) with deionized water for 3 minutes and subsequently air dried.

All µLCs, except for IQ SQIN, were provided as ready-to-use pastes. The IQ SQIN powder was mixed with the corresponding liquid to create a solid paste. The first layer of paste was placed on the occlusal surface with a brush and then fired. A second layer was added to achieve an approximate thickness of 0.5 mm and refired.

Table 1. Summary of tested ceramic materials, abbreviations, manufacturers, lot numbers, compositions, and coefficient of thermal expansion (CTE) according to manufacture information

Group Abbreviation	Ceramic Material	Manufacturers	Lot. no.	Composition	CTE ($\times10^{-6}\times K^{-1}$)
CER	ceraMotion ZR Paste Glaze transpa	Dentaurum	2310245A	N.N.	8.9
	ceraMotion 3D neutral		2211060A	SiO_2, Al_2O_3, K_2O, Na_2O, Li_2O, CaO, MgO, B_2O_3, BaO, pigment	9.1
IQS	Lustre Paste One neutral	GC Europe N.V.	2305251	N.N.	8
	IQ SQIN E−57		2306151	N.N.	8.5
MIY	MiYO trans Shade A	Chemichl	162309	SiO_2, Al_2O_3, B_2O_3, K_2O, Na_2O, CeO_2, $Sn0_2$, ZnO, CaO, Li_2O, F, pigment	7.5
	MiYO structure window		392308	SiO_2, Al_2O_3, B_2O_3, K_2O, Na_2O, CeO_2, $Sn0_2$, ZnO, P_2O_5, CaO, Li_2O, F, pigment	7.4
STR	Structure clear	Estetic ceram	251023	SiO_2, Al_2O_3, K_2O, Na_2O, Li_2O, SrO, B_2O_3, CeO_2, ZnO + pigment	9.5
VEN	Zirkon clear	Estetic ceram	271123	SiO_2, Al_2O_3, K_2O, Na_2O, CaO, B_2O_3	9.5
MZ	Zolid HT+ white	Amann Girrbach AG	172002, 02310001, 02004001	4Y-TZP	10.2

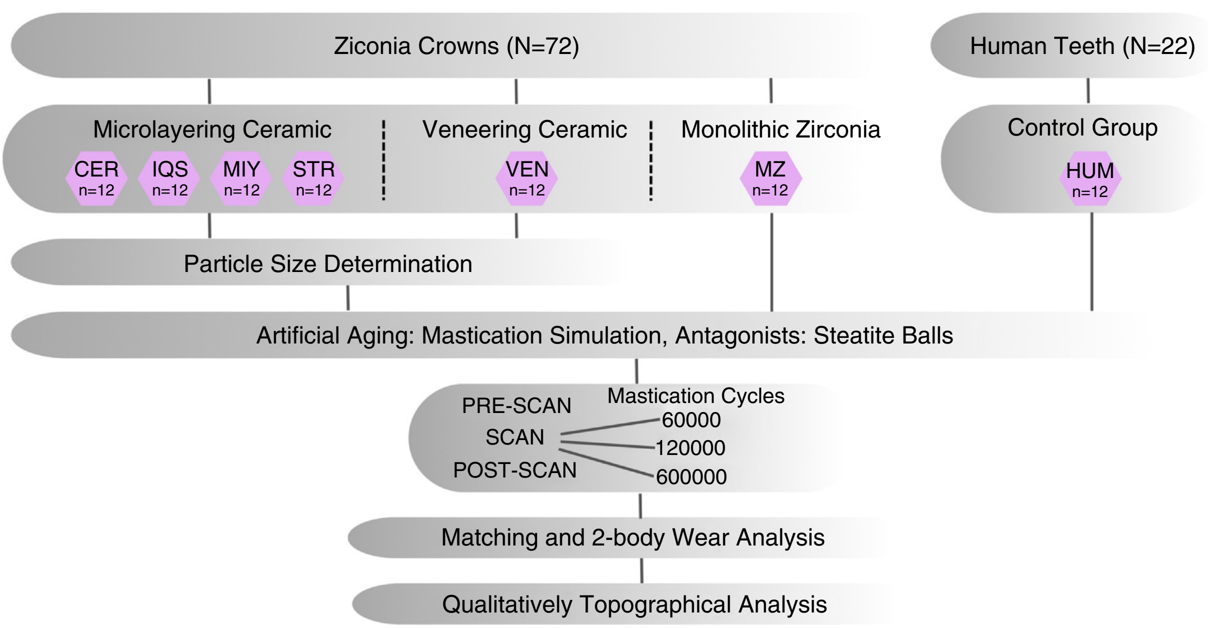

Figure 1. Study design.

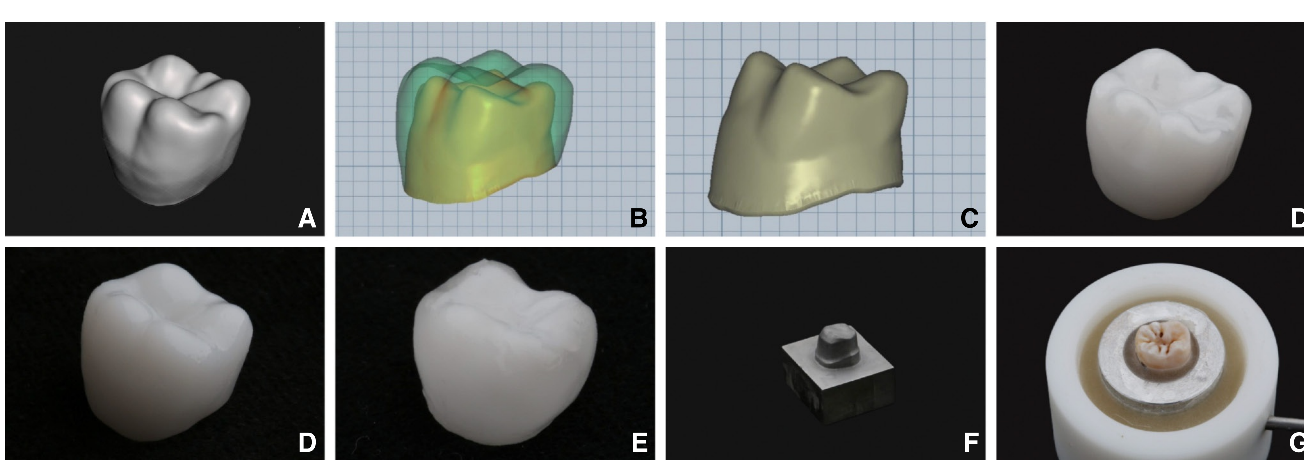

Figure 2. Photographed materials. A, 3D-structure of crown with medium cusp inclination. B, Scanned and subsequently reduced microlayered crown. C, 3D-structure of reduced crown. D, Polished crown. E, Microlayered crown. F, Veneered crown. G, Alloy abutment. H, Human tooth.

To ensure a standardized shape and size for the veneered crowns, a silicone mold (Tec C-Silicone; Orbis Dental) was created based on the morphology of the microlayered crowns.

Veneering ceramic powder was mixed with the corresponding liquid to form a slurry. After the initial firing, any deficiencies resulting from sintering shrinkage were compensated for by adding additional veneering ceramic using the mold once more, followed by a second firing. The firing schedule was conducted in a single ceramic oven (Austromat 654 press-i-dent; Dekema Dental-Keramiköfen GmbH) with 4 crowns placed on each mesh tray and adhering to the instructions provided by each ceramic manufacturer (Table 2).

µLC crowns were manually polished for approximately 2 minutes at 7000 rpm using ceramic rubber polishers (Gebr. Brasseler GmbH & Co. KG) and finished with goat-hair brushes (Orbis Dental) along with diamond polishing paste (DIA-GLACE; YETI Dentalprodukte GmbH).

Human teeth were cleaned and stored in a 0.5% chloramine solution (Chloramine-T Trihydrat; Sigma-Aldrich; CAS No: 7080–50-4) at room temperature (23 °C) for 1 week following extraction. The use of human-derived teeth for those in vitro studies had been approved by the Ethics Committee of the Ludwig-Maximilians-University Munich (LMU Munich, Germany; approval letter dated July 10, 2012). The teeth were then transferred to distilled water and refrigerated at 5 °C for up to 6 months. The roots of the human molars were embedded in circular metal specimen holders (DR-Tec GmbH) using casting resin (SCANDIQUICK; SCAN-DIA GmbH). These specimens were then stored in distilled water at 37 °C for 24 hours in an incubator (HeraCell 150; Heraeus Group). Steatite balls (SD-Mechatronik GmbH) (Ø6 mm) were also embedded in metal holders using casting resin, serving as antagonists during the mastication simulation.

Before luting, the intaglio surfaces of each crown were airborne-particle abraded for 10 seconds using 50-µm alumina at 0.05 MPa from 10 mm. The crowns were subsequently cleaned in an ultrasonic bath with deionized water for 3 minutes and then air dried. A ceramic primer (CERAMIC PRIMER PLUS; Kuraray Noritake Dental Inc) was applied and gently rubbed into the surface using a microbrush (Kuraray Noritake Dental Inc), followed by air drying with oil-free air. The crowns were bonded to cobalt chromium (Co-Cr) abutments (Ceramill Sitron, Amann Girrbach AG) with a chamfer preparation (abutment height: 5.8 mm, preparation angle: 6 degree) using a dual-polymerized resin composite (Panavia V5 Paste; Kuraray Noritake Dental Inc).[14,25] Finger pressure was applied for 30 seconds to ensure proper seating. After setting for 10 minutes, the specimens were stored in deionized water at 37 °C for 24 hours.

For the 2-body wear testing, all specimens underwent mastication simulation (CS-4; SD Mechatronik GmbH) with steatite antagonists. The crowns were positioned to achieve 3-point contact on the mesiobuccal, mediobuccal, and mesiolingual cusps, verified using occlusal foil (ArtiFol, Dr. Jean Bausch GmbH & Co. KG). This design resulted in multiple wear facets per specimen. For volumetric wear assessment, the occlusal measurement area was defined to encompass all 3 contact regions, ensuring that material loss across all wear facets was included in the quantitative analysis. The simulation parameters were set as follows: vertical stroke of 2 mm, lateral movement of 0.7 mm, and load of 50 N at a frequency of 1.3 Hz, with a total of 600 000 masticatory cycles. To simulate oral conditions, thermal loading was applied by alternating water temperature between 5 °C and 55 °C at intervals of 60 seconds. Impressions (Flexitime Fast&Scan; Kulzer GmbH) of the crowns and the antagonists were made at 4 time points: before mastication simulation, after 60 000 cycles, after 120 000 cycles, and at the conclusion of the simulation (after 600 000 cycles). The impressions were cleaned with air, sprayed with a scan spray for 2 seconds and a thickness of 3 µm (Arti-Spray BK 285; Dr. Jean Bausch GmbH & Co KG) and scanned with a laser scanner (LAS-20; SD Mechatronik GmbH) with a horizontal resolution of 20 ±5 µm and a vertical resolution of 0.8 µm. Using a 3D inspection software program

Table 2. Summary of firing parameters of ceramic materials, their number of firings, dry time, start temperature, heating rate, vacuum start and end, firing temperature, and holding time

Firing Number	Ceramic Material	Dry Time	Start Temperature	Heating Rate	Vacuum Start	Vacuum End	Firing Temperature	Holding Time
1.	ceraMotion ZR Paste Glaze transpa	8 min	450 °C	55 °C/min	450 °C	735 °C	735 °C	1 min
2.	ceraMotion 3D neutral							
1.	Lustre Paste One neutral	6 min	450 °C	45 °C/min	850 °C	850 °C	850 °C	1 min
2.	IQ SQIN E–57	8 min	400 °C	40 °C/min	780 °C	780 °C	780 °C	1 min
1.	MiYO trans Shade A	8 min	450 °C	45 °C/min	580 °C	720 °C	720 °C	1 min
2.	MiYO structure window							
1. + 2.	Structure clear	7 min	400 °C	45 °C/min	670 °C	770 °C	770 °C	1 min
1.	Zirkon clear	4 min	400 °C	45 °C/min	450 °C	780 °C	780 °C	1 min
2.						770 °C	770 °C	

(INSPECT Optical 3D; Carl Zeiss GOM Metrology GmbH), the scan data were analyzed to quantify the volume loss (mm³) after the 3 cycles of masticatory simulations. Volume loss in areas showing visible alterations was measured through an integral function, which involved overlaying the mesh structures from the prescan and postinterval scans using a Gaussian best-fit approach. For the qualitative assessment of abraded areas, all crown and antagonist surfaces were examined under a digital light microscope (VHX-S600E; KEYENCE Corp) to identify surface changes such as glaze loss, material exposure, and general wear patterns.

To determine particle size distribution for the ceramic materials, a laser particle sizer (Analysette 22; Fritsch GmbH) using laser diffraction techniques (inverse Fourier optics) was used. A spatula tip of the ceramic paste was dissolved in 3 mL of water and then added to the wet dispersion unit of the particle size analyzer. To minimize potential sources of error in measurements, 3 separate assessments were conducted for each material to obtain a mean value. The Q3(x) percentage value was calculated to facilitate comparisons among different materials by indicating what percentage of the total specimen volume consisted of particles smaller than a specified size. In this study, Q3(50), also referred to as D50, was used, signifying that 50% of the specimen volume was made up of particles smaller than this value.

The normality of data distribution within each material and mastication cycle was assessed using the Kolmogorov-Smirnov test. As the data did not follow a normal distribution, nonparametric statistical methods were used throughout. The relationship between vertical and volumetric wear for crowns and antagonists per material and aging group was evaluated using the Spearman rank correlations. For the analysis of particle size, Kruskal-Wallis and Mann-Whitney-U-tests were applied with a statistical software program (IBM SPSS Statistics, v29.0; IBM Corp) (α=.05). To assess global difference in volumetric wear (separately for crowns and antagonists) across the 7 materials and 3 measurements time points, a nonparametric ANOVA-type statistic (ATS) for longitudinal data was applied (α=.05). Pairwise post hoc comparisons were conducted with Bonferroni correction (α=.0034). The correlation between volumetric wear and particle size was assessed separately for crown and antagonists using the Kendall rank correlation coefficient with a statistical software program (R software, V4.3.1, R Core Team, www.r-project.org).

RESULTS

Strong positive correlations were found between vertical and volumetric wear for crowns and antagonists (*P*<.001). Consequently, only volumetric wear is reported. Normality was violated in 19% of crown wear groups, 33% of antagonist wear groups, and 40% of particle size groups (Table 3).

Table 3. Mean ±standard deviation of volumetric losses for all tested crowns and antagonists

Material		Volumetric Loss (mm³) Crown Mean ±SD	Antagonist Mean ±SD
60 000 masticatory cycles			
microlayered	CER	−0.1767 ±0.065	−0.0192 ±0.021
	IQS	−0.1700 ±0.070	−0.0375 ±0.024*
	MIY	−0.2383 ±0.076	−0.0292 ±0.040
	STR	−0.1367 ±0.096	−0.0608 ±0.030
veneered	VEN	−0.1308 ±0.058	−0.0183 ±0.023
polished	MZ	0 ±0	−0.0425 ±0.020
control	HUM	−0.2908 ±0.314*	0 ±0
120 000 masticatory cycles			
microlayered	CER	−0.3475 ±0.119	−0.0442 ±0.045*
	IQS	−0.3133 ±0.114*	−0.1125 ±0.076*
	MIY	−0.3892 ±0.110	−0.1258 ±0.129*
	STR	−0.2650 ±0.116	−0.1117 ±0.070
veneered	VEN	−0.2300 ±0.118	−0.1117 ±0.093*
polished	MZ	0 ±0	−0.0792 ±0.025
control	HUM	−0.3292 ±0.489*	−0.0008 ±0.003*
600 000 masticatory cycles			
microlayered	CER	−1.1050 ±0.448	−0.4292 ±0.199
	IQS	−1.1800 ±0.378	−0.5300 ±0.174
	MIY	−1.3175 ±0.419	−0.4858 ±0.250
	STR	−1.0492 ±0.429	−0.6492 ±0.123
veneered	VEN	−0.8158 ±0.319	−0.4650 ±0.152
polished	MZ	0 ±0	−0.4908 ±0.186
control	HUM	−0.7142 ±0.811*	−0.0017 ±0.006*

CER, Ceramotion One Touch 3D; HUM, human teeth; IQS, IQ SQIN; MIY, MiYO structure; MZ, monolithic zirconia; VEN, veneered zirconia.
* Deviation from normal distribution

Material type significantly influenced volumetric wear (crowns: *P*=.003; antagonists: *P*<.001), as did the number of masticatory cycles (both *P*<.001) (Table 4). The progression of wear differed among material groups for both crowns and antagonists (*P*<.001) (Fig. 3). Among the crown materials, VEN showed lower loss than MIY (*P*<.001). MIY showed higher material loss than STR (*P*=.002). The remaining µLCs fell within the same range of values (*P*=.088 to.245). HUM likewise fell within the same range of values as CER (*P*=.191), STR (*P*=.642), IQS (*P*=.268), and VEN (*P*=.950). All tested materials showed more wear than HUM (*P*<.001). MZ experienced such minimal material losses that they were undetectable by the scanner and omitted from the data analysis. On the antagonists, STR showed higher loss than CER (*P*=.001). HUM demonstrated the lowest wear (*P*<.001), followed by CER (*P*<.001). All tested materials showed higher wear increase than HUM (*P*<.001). MZ experienced such minimal material losses that they were undetectable by the scanner and omitted from the data analysis. Generally, the material loss increased with the number of masticatory cycles. This trend was consistent across all groups, except for the MZ crowns and the antagonists against HUM, which exhibited no wear.

After 600 000 masticatory cycles the digital light microscope revealed exposure of the zirconia framework as

Table 4. Nonparametric ANOVA-type statistic (ATS) for longitudinal data with a global difference in volumetric wear, separately for crowns and antagonists

Parameters	Crowns			Antagonists		
	Rank Means	Nobs	Relative Treatment Effect	Rank Means	Nobs	Relative Treatment Effect
CER	101	36	0.463	135	36	0.534
IQS	103	36	0.472	103	36	0.408
MIY	85	36	0.391	115	36	0.455
STR	115	36	0.529	91	36	0.359
VEN	125	36	0.578	115	36	0.456
MZ	could not be quantified			107	36	0.423
HUM	123	36	0.567	219	36	0.865
60 000 cycles	161	72	0.743	179	84	0.710
120 000 cycles	122	72	0.560	136	84	0.537
600 000 cycles	43	72	0.197	64	84	0.253
CER * 60 000 cycles	162	12	0.748	191	12	0.755
CER * 120 000 cycles	104	12	0.478	164	12	0.648
CER * 600 000 cycles	36	12	0.164	50	12	0.198
IQS * 60 000 cycles	165	12	0.759	168	12	0.666
IQS * 120 000 cycles	113	12	0.523	107	12	0.422
IQS * 600 000 cycles	30	12	0.135	35	12	0.136
MIY * 60 000 cycles	136	12	0.629	184	12	0.727
MIY * 120 000 cycles	94	12	0.433	116	12	0.459
MIY * 600 000 cycles	24	12	0.111	46	12	0.179
STR * 60 000 cycles	179	12	0.825	141	12	0.557
STR * 120 000 cycles	129	12	0.594	113	12	0.444
STR * 600 000 cycles	37	12	0.168	19	12	0.075
VEN * 60 000 cycles	182	12	0.842	190	12	0.753
VEN * 120 000 cycles	144	12	0.665	113	12	0.447
VEN * 600 000 cycles	49	12	0.226	43	12	0.168
MZ * 60 000 cycles	could not be quantified			161	12	0.638
MZ * 120 000 cycles				120	12	0.475
MZ * 600 000 cycles				40	12	0.158
HUM * 60 000 cycles	141	12	12	221	12	0.879
HUM * 120 000 cycles	145	12	12	218	12	0.862
HUM * 600 000 cycles	82	12	12	217	12	0.860

CER, Ceramotion One Touch 3D; HUM, human teeth; IQS, IQ SQIN; MIY, MiYO structure; MZ, monolithic zirconia; VEN, veneered zirconia.

follows: MIY (1 of 12), STR (1 of 12), CER (7 of 12). On IQS and VEN the zirconia framework was not exposed (Fig. 4). MIY, followed by IQS, CER, STR, and VEN, exhibited the lowest particle size (Table 5) ($P<.001$). A negative correlation was found between the particle size and volumetric loss of the crowns ($P<.001$).

DISCUSSION

The aim of this in vitro study was to investigate the wear behavior of µLCs in relation to their particle size. The null hypothesis that no significant differences would be found in the wear behavior of the investigated groups (microlayered and veneered zirconia crowns, monolithic

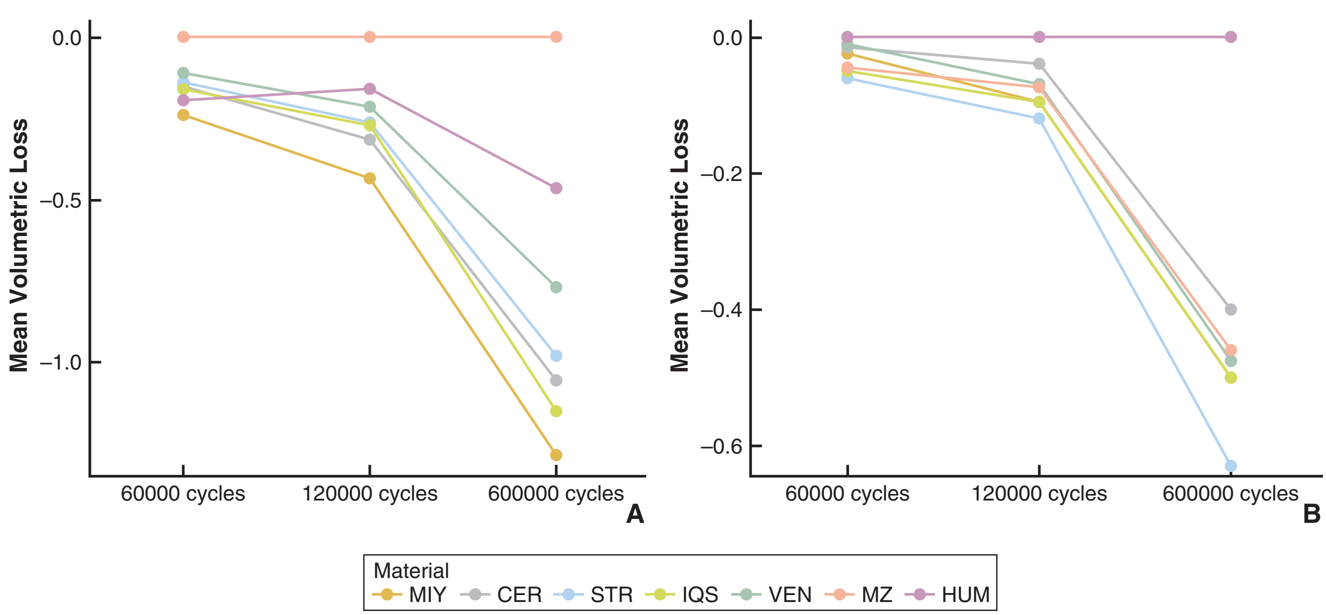

Figure 3. Median wear by masticatory cycles and materials. A, Crowns. B, Antagonists.

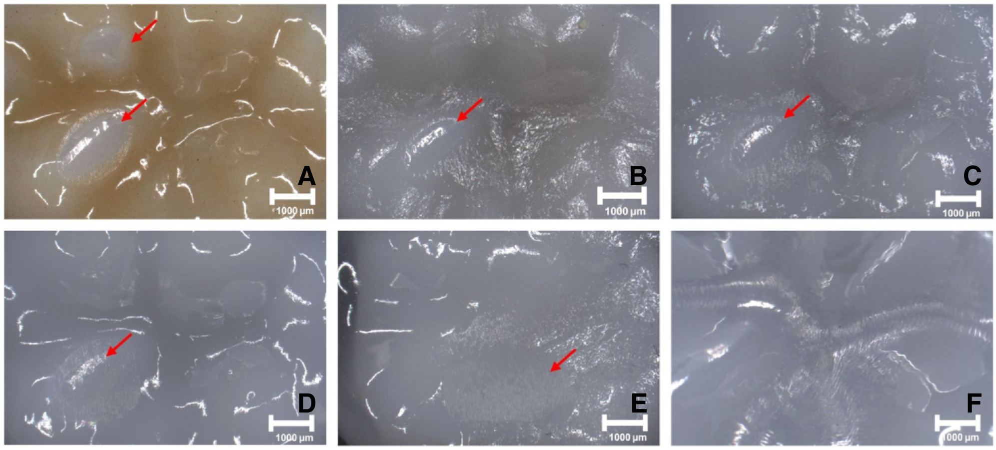

Figure 4. Exposure of rough zirconia framework, (Original magnification, ×30), after 600 000 masticatory cycles on A, MIY. B, CER. C, STR. D, wear area of IQS. E, wear area of VEN and F, no visible wear of MZ. *Red arrows* show wear areas. CER, Ceramotion One Touch 3D; IQS, IQ SQIN; MIY, MiYO structure; MZ, monolithic zirconia; VEN, veneered zirconia.

Table 5. Mean ±standard deviation of particle size for microlayering and veneering ceramics

Material		Particle Size (µm)
		Mean ±SD
Microlayered	CER	13.8 ±0[c]
	IQS	12.0 ±0[b]
	MIY	9.8 ±0*[a]
	STR	15.5 ±0[d]
Veneered	VEN	24.1 ±0*[e]

Lowercase letters indicate differences between material groups
Different letters indicate significant difference (P<.05)
CER, Ceramotion One Touch 3D; IQS, IQ SQIN; MIY, MiYO structure; MZ, monolithic zirconia; VEN, veneered zirconia.
* Deviation from normal distribution

zirconia, and human teeth) was rejected as material-dependent wear patterns were found.

A correlation was observed between particle size and volumetric material loss on the VEN and µLCs, indicating that material loss increased with decreasing particle size (Fig. 5). Rosentritt et al[7] reported contradictory findings, noting higher material loss for veneered zirconia than for painted zirconia. However, lower layer thicknesses were applied in that study. A correlation between layer thickness and wear values has also been documented,[7] along with a significant effect of the number of masticatory cycles on wear.[31] In the present study as well, an increasing material loss was observed with an increasing number of masticatory cycles. Microscopic evaluation (Fig. 4) after 600 000 masticatory cycles (equivalent to 2.5 years in vivo)[34,35] revealed partial exposure of the zirconia framework in some µLCs groups. This was consistent with a previous report showing glaze material removal after 120 000 cycles.[31] These findings highlight the dynamic nature of wear and reinforce the need for long-term data to better understand when and how surface degradation occurs.

Clinically, the ability of µLCs and VEN to adapt through wear may help even distribution of contacts between maxillary and mandibular teeth and reduce functional stress, an advantage over polished surfaces.[32] VEN demonstrated wear patterns most similar to those of natural enamel, suggesting a more physiological response. Contrary to expectations and previous studies,[2,32] MZ did not reduce antagonist wear compared with VEN and µLCs. Among all groups, antagonists opposing HUM showed the lowest wear values, consistent with the lower hardness (330 HV)[27] and initial roughness (0.9 ±0.2 µm)[17] compared with steatite (680 HV[27]; initial roughness: 1.7 ±0.2 µm[17]). Unexpectedly, STR caused higher material loss on steatite than CER.

After the mastication simulation, no measurable wear was observed on MZ, which was consistent with findings from previous studies.[14–16] The high hardness and wear resistance of MZ likely contributed to this outcome. Although no visible material loss occurred on the polished zirconia crowns, wear was still observed on the opposing steatite antagonists. This phenomenon can be attributed to the significant hardness difference between the materials, where the harder zirconia surface may induce wear on the softer antagonist without itself being visibly affected. Clinically, restorative materials that exhibit wear behavior similar to that of human enamel are preferable, particularly in single-tooth restorations within a completely dentate arch of natural, unrestored teeth. Matching the physiological wear pattern supports the preservation of occlusal balance and vertical dimension.

Another factor potentially increasing wear was the CTE (Table 1). In this study, IQS and MIY did not adhere to the recommended CTE proximity, exhibiting values more than 10% lower than those of the zirconia framework. This mismatch may have contributed to the

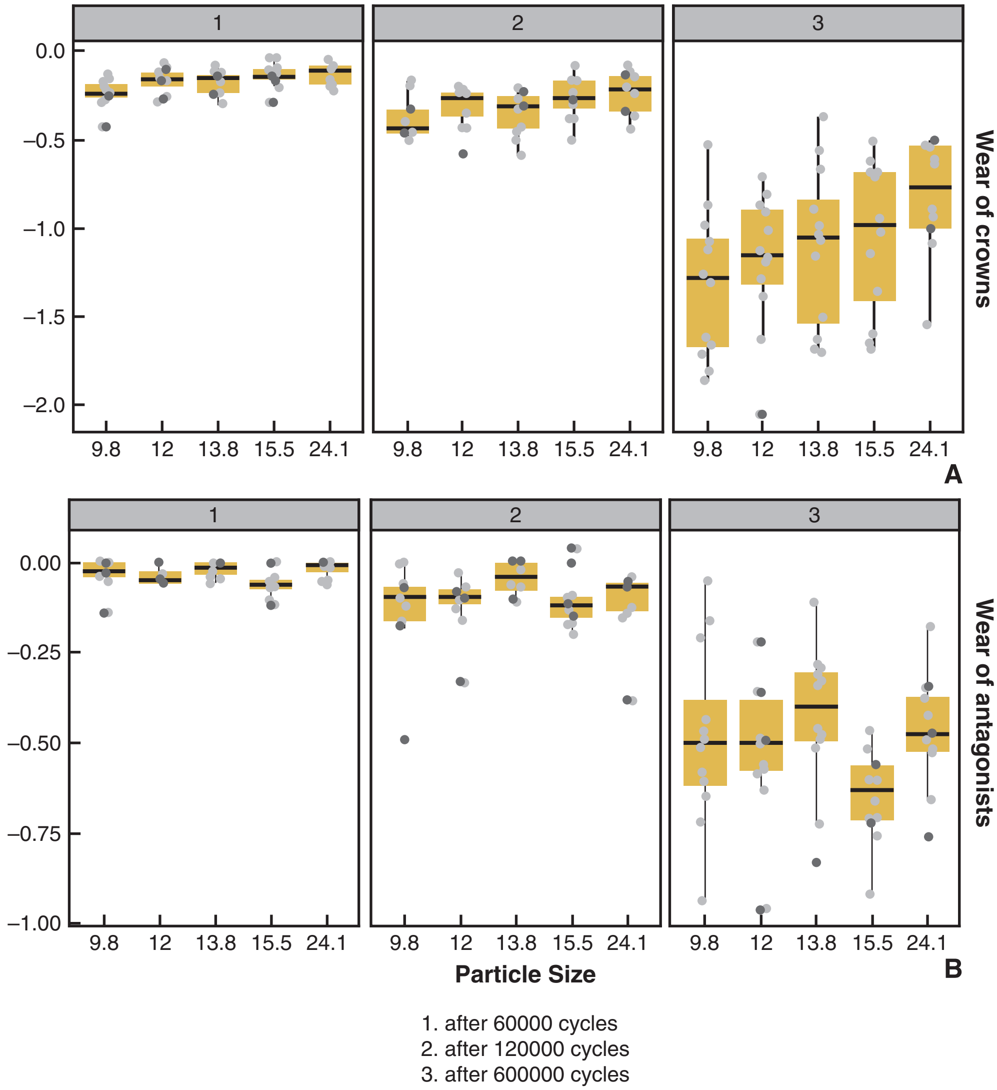

Figure 5. Relationship between wear and particle size. A, Crowns. B, Antagonists.

higher wear observed in these groups. According to established principles, a slight CTE mismatch favoring compressive stress in the veneer is desirable, as ceramics are stronger under compression.[11,12]

In this study, to enhance clinical relevance, anatomic crowns designs were used, as flatter geometries have been shown to exaggerate wear.[22,39] Cusp inclination has also been reported to influence wear, with steeper inclines linked to higher material loss.[25] Steatite balls were selected as antagonist material because of their standardized geometry and established use in wear studies.[27,32] As a multicomponent semiporous crystalline ceramic,[26] steatite differs from human enamel in hardness and surface roughness.[17,27] Natural human molars were used to mimic physiological behavior, but this introduced variability, as reflected in the wider data spread. To mitigate geometric discrepancies, teeth with low cusp height and crown dimensions resembling those of the standardized crowns were selected. The choice of abutment material has also been reported to impact wear.[25] Co-Cr was selected based on its mechanical reliability and comparability with enamel.[25] Compared with natural enamel abutments, Co-Cr offers consistent quality and avoids variations in morphology and composition, providing a more standardized test design.[25] The mastication force (50 N) and frequency (1.3 Hz) used in this study reflected average clinical values, ensuring comparability with other studies.[17,25,29,30] Additionally, thermocycling between 5 °C and 55 °C introduced aging effects, while water flushing the test chambers helped eliminate wear debris.[17]

Limitations of the study included the in vitro design. Clinical factors such as salivary flow, patient-specific chewing forces, parafunctional habits, and the presence of enamel antagonists were not modeled. These parameters are essential for understanding real-world material behavior and should be considered in future clinical trials.

CONCLUSIONS

Based on the findings of this in vitro study, the following conclusions were drawn:

1. µLC and VEN exhibited material loss during mastication, which aligned more closely with the wear behavior of HUM than with that of MZ, which showed little to no wear.
2. Among the tested materials, VEN showed wear most similar to HUM, indicating a more physiological response under simulated mastication conditions.
3. However, as wear increased with the number of masticatory cycles, µLC could be worn away over time.

REFERENCES

1. Rauch A, Schrock A, Schierz O, Hahnel S. Material selection for tooth-supported single crowns-A survey among dentists in Germany. *Clin Oral Investig*. 2021;25:283–293.
2. Zhang F, Spies BC, Vleugels J, et al. High-translucent yttria-stabilized zirconia ceramics are wear-resistant and antagonist-friendly. *Dent Mater*. 2019;35:1776–1790.
3. Bruhnke M, Awwad Y, Muller WD, Beuer F, Schmidt F. Mechanical properties of new generations of monolithic, multi-layered zirconia. *Materials (Basel)*. 2022;16:276.
4. Piconi C, Maccauro G. Zirconia as a ceramic biomaterial. *Biomaterials*. 1999;20:1–25.
5. Denry I, Kelly JR. State of the art of zirconia for dental applications. *Dent Mater*. 2008;24:299–307.
6. Kontonasaki E, Rigos AE, Ilia C, Istantsos T. Monolithic zirconia: An update to current knowledge. optical properties, wear, and clinical performance. *Dent J (Basel)*. 2019;7:90.
7. Rosentritt M, Bollin D, Schmidt MB, Rauch A. Stability and wear of zirconia crowns with micro-layering. *J Dent*. 2023;135:104560.
8. Zhang F, Reveron H, Spies BC, Van Meerbeek B, Chevalier J. Trade-off between fracture resistance and translucency of zirconia and lithium-disilicate glass ceramics for monolithic restorations. *Acta Biomater*. 2019;91:24–34.
9. Zhang Y, Lawn BR. Novel zirconia materials in dentistry. *J Dent Res*. 2018;97:140–147.
10. Limones A, Molinero-Mourelle P, Azevedo L, Romeo-Rubio M, Correia A, Gomez-Polo M. Zirconia-ceramic versus metal-ceramic posterior multiunit tooth-supported fixed dental prostheses: A systematic review and meta-analysis of randomized controlled trials. *J Am Dent Assoc*. 2020;151:230–238 e7.
11. Fischer J, Stawarczyk B, Tomic M, Strub JR, Hammerle CH. Effect of thermal misfit between different veneering ceramics and zirconia frameworks on in vitro fracture load of single crowns. *Dent Mater J*. 2007;26:766–772.
12. Coffey JP, Anusavice KJ, DeHoff PH, Lee RB, Hojjatie B. Influence of contraction mismatch and cooling rate on flexural failure of PFM systems. *J Dent Res*. 1988;67:61–65.
13. Mainjot AK, Najjar A, Jakubowicz-Kohen BD, Sadoun MJ. Influence of thermal expansion mismatch on residual stress profile in veneering ceramic layered on zirconia: Measurement by hole-drilling. *Dent Mater*. 2015;31:1142–1149.
14. Hoffmann M, Mayinger F, Stawarczyk B. Influence of different surface finishing procedures of strength-gradient multilayered zirconia crowns on two-body wear and fracture load: Lithium silicate or leucite glazing versus polishing? *J Mech Behav Biomed Mater*. 2023;150:106307.
15. Janyavula S, Lawson N, Cakir D, Beck P, Ramp LC, Burgess JO. The wear of polished and glazed zirconia against enamel. *J Prosthet Dent*. 2013;109:22–29.
16. Bai Y, Zhao J, Si W, Wang X. Two-body wear performance of dental colored zirconia after different surface treatments. *J Prosthet Dent*. 2016;116:584–590.
17. Preis V, Behr M, Kolbeck C, Hahnel S, Handel G, Rosentritt M. Wear performance of substructure ceramics and veneering porcelains. *Dent Mater*. 2011;27:796–804.
18. Leon Velastegui M, Montiel-Company JM, Agustin-Panadero R, Fons-Badal C, Sola-Ruiz MF. Enamel wear of antagonist tooth caused by dental ceramics: Systematic review and meta-analysis. *J Clin Med*. 2022;11:6547.
19. Rosentritt M, Preis V, Behr M, Hahnel S, Handel G, Kolbeck C. Two-body wear of dental porcelain and substructure oxide ceramics. *Clin Oral Investig*. 2012;16:935–943.
20. Kim SK, Kim KN, Chang IT, Heo SJ. A study of the effects of chewing patterns on occlusal wear. *J Oral Rehabil*. 2001;28:1048–1055.
21. Mair LH, Stolarski TA, Vowles RW, Lloyd CH. Wear: Mechanisms, manifestations and measurement. Report of a workshop. *J Dent*. 1996;24:141–148.
22. Heintze SD, Cavalleri A, Forjanic M, Zellweger G, Rousson V. Wear of ceramic and antagonist–A systematic evaluation of influencing factors in vitro. *Dent Mater*. 2008;24:433–449.
23. Albashaireh ZS, Ghazal M, Kern M. Two-body wear of different ceramic materials opposed to zirconia ceramic. *J Prosthet Dent*. 2010;104:105–113.
24. Lambrechts P, Braem M, Vuylsteke-Wauters M, Vanherle G. Quantitative in vivo wear of human enamel. *J Dent Res*. 1989;68:1752–1754.
25. Schmeiser F, Arbogast F, Ruppel H, Mayinger F, Reymus M, Stawarczyk B. Methodology investigation: Impact of crown geometry, crown, abutment and antagonist material and thermal loading on the two-body wear of dental materials. *Dent Mater*. 2022;38:266–280.
26. Arranz J.G.A., R A. New philosophies in ceramic layering. GC get connected.
27. Shortall AC, Hu XQ, Marquis PM. Potential countersample materials for in vitro simulation wear testing. *Dent Mater*. 2002;18:246–254.
28. Rosentritt M, Behr M, Gebhard R, Handel G. Influence of stress simulation parameters on the fracture strength of all-ceramic fixed-partial dentures. *Dent Mater*. 2006;22:176–182.
29. Rosentritt M, Behr M, van der Zel JM, Feilzer AJ. Approach for valuating the influence of laboratory simulation. *Dent Mater*. 2009;25:348–352.
30. Rosentritt M, Siavikis G, Behr M, Kolbeck C, Handel G. Approach for valuating the significance of laboratory simulation. *J Dent*. 2008;36:1048–1053.
31. Stawarczyk B, Ozcan M, Schmutz F, Trottmann A, Roos M, Hammerle CH. Two-body wear of monolithic, veneered and glazed zirconia and their corresponding enamel antagonists. *Acta Odontol Scand*. 2013;71:102–112.
32. Preis V, Grumser K, Schneider-Feyrer S, Behr M, Rosentritt M. Cycle-dependent in vitro wear performance of dental ceramics after clinical surface treatments. *J Mech Behav Biomed Mater*. 2016;53:49–58.
33. Wimmer T, Hoffmann AMS, Eichberger M, Schmidlin PR, Stawarczyk B. Two-body wear rate of PEEK, CAD/CAM resin composite and PMMA: Effect of specimens geometries, antagonist materials and test set-up configuration. *Dent Mater*. 2016;31:e127–e136.
34. Sakaguchi RL, Douglas WH, DeLong R, Pintado MR. The wear of a posterior composite in an artificial mouth: A clinical correlation. *Dent Mater*. 1986;2:235–240.
35. DeLong R, Sakaguchi RL, Douglas WH, Pintado MR. The wear of dental amalgam in an artificial mouth: A clinical correlation. *Dent Mater*. 1985;1:238–242.

Corresponding author:
Prof Bogna Stawarczyk
Department of Prosthetic Dentistry
University Hospital
LMU Munich
Goethestraße 70
Munich 80336
GERMANY
Email: bogna.stawarczyk@med.uni-muenchen.de

https://doi.org/10.1016/j.prosdent.2025.10.055