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**Digitale Workflows in der restaurativen Zahnmedizin:
Einfluss von Abformtechnik, Materialauswahl,
Restaurationsdesign und Schichtstärke**

Habilitationsschrift

zur Erlangung der Venia Legendi für das
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***„Versprich mir, dass du immer daran denkst: Du bist mutiger, als du glaubst,
stärker, als du aussiehst und schlauer, als du denkst.“***

A. A. Milne

In Dankbarkeit meiner Familie

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1. Abkürzungsverzeichnis

3D	Dreidimensional
3Y-TZP	3 mol% yttriumoxidstabilisiertes tetragonales Zirkoniumoxid-Polykristall
4Y-PSZ	4 mol% yttriumoxidteilstabilisiertes Zirkoniumoxid
5Y-PSZ	5 mol% yttriumoxidteilstabilisiertes Zirkoniumoxid
ANOVA	Analysis of Variance
BSE	Back Scattered Electrons
CAD	Computer-Aided Design
CAM	Computer-Aided Manufacturing
CEREC	Chairside Economical Restoration of Esthetic Ceramics
CT	Computertomographie
IOS	Intraoralscanner
IPS	Ivoclar Pressed System (Markenbezug, z. B. IPS e.max)
ISO	International Organization for Standardization
KI	Künstliche Intelligenz
LDS	Lithiumdisilikat
MDP	10-Methacryloyloxydecyl-dihydrogenphosphat
OBJ	Objekt (3D-Datenformat)
PG	Photogrammetrie
PIC	Precise Implants Capture (Herstellerbezug)
PICN	Polymer-Infiltrated Ceramic Network
PLY	Polygon File Format (3D-Datenformat)
PMMA	Polymethylmethacrylat
REM	Rasterelektronenmikroskop
SLM	Selective Laser Melting
STL	Standard Tessellation Language (3D-Datenformat)
TRIOS	Markenname Intraoralscanner (3Shape)

2. Beitragende Originalarbeiten

Im Rahmen der vorliegenden Habilitationsschrift werden zentrale Einflussfaktoren des digitalen Workflows untersucht und mögliche Optimierungsstrategien evaluiert. Die Grundlage bilden fünf Publikationen, die sich mit im Computer-Aided-Design/Computer-Aided-Manufacturing-(CAD/CAM)-Workflow hergestellten Einzelzahnrestorationen sowie Titan-Gerüsten befassen. Untersucht wurde die vertikale Passung implantatgetragener Ganzkiefergerüste, die mit unterschiedlichen digitalen Abformmethoden hergestellt wurden, sowie der Einfluss des Keramikmaterials, Restaurationsdesigns und der Schichtstärke auf die Bruchfestigkeit von keramischen Einzelzahnrestorationen (Kronen, Teilkronen und okklusale Veneers). Ergänzend erfolgte eine fraktographische Analyse rissbildender Proben nach thermodynamischer Ermüdung sowie eine theoretische Modellierung zur Lebensdauereinschätzung, um praxisrelevante Empfehlungen für die klinische Anwendung abzuleiten.

Die Einleitung ermöglicht zunächst einen Überblick über die Entwicklung und den aktuellen Stand digitaler Workflows in der restaurativen Zahnmedizin. Im Anschluss werden bestehende Limitationen des digitalen Workflows und offene Forschungsfragen aufgezeigt sowie die wissenschaftlichen Fragestellungen der vorliegenden Arbeit erläutert. Im Abschnitt „Originalarbeiten“ werden Hintergründe und die Methodik der einzelnen Studien dargestellt, ihre Ergebnisse zusammengefasst und diskutiert. Abschließend erfolgt eine Gesamtbetrachtung der gewonnenen Erkenntnisse und deren Einordnung in den wissenschaftlichen Kontext sowie ein Ausblick auf zukünftige Perspektiven und Forschungsansätze.

Originalarbeit 1:

Conejo J, Ordway M, Yoo TH, Fraiman HP, Atria PJ, Retana L, Blatz MB, Prott LS.

In vitro comparison between complete-arch implant supported frameworks using intraoral scanning and photogrammetry.

Int J Prosthodont. 2025 May 20;0(0):1-17. doi: 10.11607/ijp.9363.

Originalarbeit 2:

Prott LS, Spitznagel FA, Bonfante EA, Malassa MA, Gierthmuehlen PC.

Monolithic zirconia crowns: Effect of thickness reduction on fatigue behavior and failure load.

J Adv Prosthodont. 2021 Oct;13(5):269-280. doi: 10.4047/jap.2021.13.5.269.

Originalarbeit 3:

Spitznagel FA, Prott LS (geteilte Erstautorenschaft), Hoppe JS, Manitzkaia T, Blatz MB, Zhang Y, Langner R, Gierthmuehlen PC.

Minimally invasive CAD/CAM lithium disilicate partial-coverage restorations show superior in vitro fatigue performance than single crowns.

J Esthet Restor Dent. 2024 Jan;36(1):94-106. doi: 10.1111/jerd.13169.

Originalarbeit 4:

Prott LS, Harlaß M, Marksteiner A, Spitznagel FA, Langner R, Zhang Y, Blatz MB, Gierthmuehlen PC.

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

J Prosthodont Res. 2025 Sep 3. Online ahead of print. doi: 10.2186/jpr.JPR_D_25_00068.

Originalarbeit 5:

Prott LS, Gierthmuehlen PC, Blatz MB, Zhang Y.

Failure prediction of lithium disilicate and composition-gradient multilayered zirconia occlusal veneers: A fractographic and theoretical analysis.

Materials (Basel). 2025 Sep 12;18(18):4287. doi: 10.3390/ma18184287.

3. Einleitung

3.1 Wandel der restaurativen Zahnmedizin durch Digitalisierung

Die restaurative Zahnmedizin hat in den vergangenen vier Jahrzehnten durch den fortschreitenden Einsatz digitaler Technologien einen tiefgreifenden Wandel erfahren [1]. Wegweisend waren insbesondere die Arbeiten von François Duret und Werner Mörmann [2, 3]. Bereits 1971 entwickelte François Duret erste Konzepte zur digitalen Erfassung der Zahnhartsubstanz und zur computergestützten Herstellung dentaler Restaurationen [4]. 1983 präsentierte er ein erstes vollständiges CAD/CAM-System für die zahnmedizinische Anwendung [5]. Parallel dazu stellte Werner Mörmann in Zusammenarbeit mit Marco Brandestini 1985 das Chairside Economical Restoration of Esthetic Ceramics (CEREC)-System vor, welches die direkte computergestützte Herstellung von Keramikrestaurationen in der Zahnarztpraxis ermöglichte [3].

In den Folgejahren erlebte die digitale Zahnmedizin eine kontinuierliche Weiterentwicklung. Die Genauigkeit intraoraler Scanner wurde sukzessive verbessert und die CAD-Software zur dreidimensionalen (3D) Konstruktion von Restaurationen gewann sowohl an Leistungsfähigkeit als auch an Benutzerfreundlichkeit [6]. Fortschritte in der Werkstoffwissenschaft ermöglichten die Entwicklung einer breiten Palette CAD/CAM-kompatibler Materialien, die heute von feldspatbasierten Keramiken bis hin zu hochfesten Lithiumdisilikaten (LDS) und Zirkoniumoxidvarianten reichen [7]. Darüber hinaus ergänzen additive Fertigungstechniken (3D-Druck) die subtraktive Bearbeitung und eröffnen neue Möglichkeiten für die Herstellung komplexer Restaurationen und Prototypen [8].

Das Konzept des digitalen Patienten zielt darauf ab, durch die Integration von intraoralen Scans, Gesichtsscans, Kieferbewegungsanalysen sowie Computertomographie (CT)-Aufnahmen ein vollständiges virtuelles 3D-Modell des Patienten zu erzeugen [9]. Dieses bildet die Grundlage für eine ästhetisch orientierte Planung, eine exakte Implantatpositionierung sowie für ein präzises Design und Herstellen der Restaurationen [10]. KI (Künstliche Intelligenz)-basierte Techniken ermöglichen dabei die automatische Ausrichtung und Überlagerung von Gesichts- und intraoralen Scans. In Intraoralscannern (IOS) kommen KI-Algorithmen zum Beispiel für die automatische Entfernung von Artefakten oder die Zuordnung von Scanbodies

mit den zugehörigen CAD-Objekten zum Einsatz [11]. Angesichts der stetigen Weiterentwicklungen im Bereich der KI und robotergestützten Zahnmedizin ist zu erwarten, dass die Digitalisierung die restaurative Zahnmedizin auch in den kommenden Jahrzehnten weiter transformieren und automatisierte, vollständig vernetzte Behandlungsabläufe ermöglichen wird [12].

Die Nutzung digitaler Technologien in Zahnarztpraxen variiert jedoch erheblich und wird maßgeblich durch die Technologiebereitschaft des zahnärztlichen Fachpersonals bestimmt. Aktuelle Umfragen in Zahnarztpraxen in Deutschland und der Schweiz ergaben, dass 44 % der allgemein Zahnärztlichen Praxen digitale Workflows nutzen. Von diesen verfügt die Mehrheit (55 %) über eine eigene Fräsmaschine, während Facescanner mit lediglich 3-9 % deutlich seltener genutzt werden [1]. Zwei Drittel der befragten deutschen Zahnärzte nutzen IOS, jedoch in den meisten Fällen nur für die Anfertigung kleinerer Restaurationen [13]. In größeren Städten und Mehrbehandlerpraxen war die Implementierung digitaler Workflows höher. Zudem wenden jüngere Zahnärzte digitale Technologien häufiger an als ältere Kollegen [1]. Finanzielle Barrieren sowie eingeschränkte Trainingsmöglichkeiten stellen wesentliche Hindernisse für die Anwendung digitaler Technologien dar [14]. 60 % der befragten Zahnärzte gaben an, innerhalb der nächsten fünf Jahre weitere Digitalisierungsmaßnahmen in ihrer Praxis zu planen, was einen deutlichen Trend von analogen zu digitalen Prozessen zeigt [13].

3.2 Konventioneller versus digitaler Workflow

3.2.1 Vor- und Nachteile

Der konventionelle Workflow umfasst die klassische Abformung mit elastomeren Materialien, die Herstellung von Gipsmodellen, das manuelle Designen der Restaurationen sowie die anschließende Fertigung mittels Press- oder Gusstechnik [15]. Trotz seiner langjährigen Etablierung ist der konventionelle Workflow aufgrund der im Vergleich zum digitalen Workflow größeren Anzahl manueller Arbeitsschritte zeitintensiver und fehleranfälliger [16, 17]. Der digitale Workflow setzt sich zusammen aus der optischen Erfassung der präparierten Zähne oder Implantate mittels IOS, dem virtuellen Designen (CAD) sowie der maschinellen Fertigung (CAM) der Restaurationen [18]. Er eröffnet Möglichkeiten, die im konventionellen Workflow

erschwert oder nicht durchführbar sind, wie Ausschneide- und Analysefunktionen, sowie eine erleichterte Wiederholbarkeit der Abformung oder Fräsprozesse. Weitere Vorteile sind die Möglichkeit der Datenfusion, die digitale Archivierbarkeit und die Chairside-Option (Behandlung in einer Sitzung) [19]. Aufgrund des höheren Komforts und der zeitlichen Effizienz wird die digitale Abformung häufig von Patienten bevorzugt. Dagegen kann die Verwendung konventioneller Abformmaterialien durch das Hervorrufen von Würgereflexen die Patientenakzeptanz negativ beeinflussen [17]. Herausforderungen in der Anwendung digitaler Workflows bestehen vor allem in der Lernkurve des Anwenders, der Erfassung subgingivaler Bereiche sowie beim Scan größerer Strecken oder zahnloser Kiefer [20].

3.2.2 Genauigkeit

Entsprechend der *International Organization of Standardization* (ISO) 5725-1 [21] wird die Genauigkeit einer Abformung über die Parameter *Richtigkeit* und *Präzision* definiert. Dabei gibt die Richtigkeit an, wie nah ein Messergebnis an einem Referenzwert liegt. Die Präzision beschreibt die Wiederholbarkeit einer Messung. Eine digitale Abformung kann präzise, aber nicht richtig sein, wenn ein systemischer Fehler vorliegt, oder richtig, aber wenig präzise, wenn zufällige Abweichungen auftreten. Für ein optimales klinisches Ergebnis ist entscheidend, dass sowohl die Richtigkeit als auch die Präzision hoch sind [22].

Die Genauigkeit einer konventionellen Abformung variiert *in vitro* je nach Größe und Beschaffenheit des Abformbereichs zwischen $9,8 \pm 0,9$ und $16,2 \pm 1,6$ μm (Richtigkeit) sowie $7,0 \pm 1,4$ und $12 \pm 4,2$ μm (Präzision) [23]. Digitale Abformungen von Einzelzahnversorgungen zeigen *in vitro* ebenfalls eine hohe Genauigkeit mit Richtigkeitswerten zwischen 18,7 und 41,7 μm sowie mit Präzisionswerten zwischen 8,3 und 24,7 μm an der Präparationsoberfläche [24]. Höher ist die Fehleranfälligkeit in vollbezahnten, teilbezahnten oder zahnlosen Kiefern [25, 26]. Für bezahnte und zahnlose Ganzkieferabformungen wurden *in vitro* Abweichungen bis zu 150 μm (bezahnt) und bis zu 193 μm (zahnlos) beschrieben [27]. Abweichungen zwischen 156 ± 8 μm (Trios 3) und 903 ± 49 μm (Planscan) wurden beim Scan einer vollbezahnten Kadavermaxilla festgestellt [26]. In einer kürzlich publizierten *in vitro* Studie wurde die Genauigkeit digitaler bezahnter Ganzkieferabformungen mit verschiedenen IOS

(Aerolscan 3, Emerald Green, iTero 5D, Medit i900, Primescan 2 und Trios 5) mit jener von konventionellen Abformungen verglichen. Obwohl alle digitalen Abformungen als klinisch akzeptabel eingestuft wurden (Richtigkeit: 30,5-95 μm , Präzision: 18,7-59,6 μm), zeigte die konventionelle Abformung eine höhere Genauigkeit [28]. Auch für den Scan zahnloser Kiefer beschreiben aktuelle in vitro Studien zwar häufig klinisch akzeptable Abweichungen [29], klinische Studien zeigen jedoch, dass die digitale Abformung des zahnlosen Ganzkiefers Limitationen aufweist mit Unterschieden von bis zu 185 ± 81 μm zwischen konventionellen und digitalen Abformungen [30]. Ein Verblocken der Abformpfosten bei implantatgetragenen Restaurationen [31] sowie die Nutzung präfabrizierter Referenzpunkte [32] kann die Genauigkeit zahnloser Ganzkieferscans verbessern. Darüber hinaus sind ein optimaler Scanpfad, eine ausreichende Überlappung der Einzelaufnahmen, kontrollierte Umgebungsbedingungen (Temperatur, Luftfeuchtigkeit, Licht) sowie die Erfahrung des Behandlers im Umgang mit dem IOS entscheidend, um die Genauigkeit digitaler Ganzkieferabformungen zu maximieren [20, 33]. Auch Soft- und Hardware-Updates haben einen signifikanten Einfluss auf die Genauigkeit einer digitalen Abformung [34].

Vor allem im Bereich implantatgetragener Ganzkieferversorgung ist die Evidenzlage hinsichtlich der Genauigkeit von digitalen im Vergleich zu konventionellen Workflows bislang uneinheitlich [35]. In der Literatur finden sich Studien, die sowohl eine höhere [32, 36] als auch geringere [37] Genauigkeit von digitalen im Vergleich zu konventionellen Verfahren beschreiben. In einer retrospektiven klinischen Analyse wurden durchschnittliche Abweichungen von 88,24 μm zwischen digitalen Abformungen und digitalisierten Modellen konventioneller Implantatabformungen festgestellt, die als klinisch akzeptabel bewertet wurden [38]. Dagegen berichtet eine kürzlich publizierte Studie, die den gesamten digitalen Workflow bewertete, von signifikanten Passungengenauigkeiten der implantatgetragenen Gerüste durch die Summation von potentiellen Scan-, Design- oder Herstellungsfehlern [35]. Eine Übersichtsarbeit zeigte ebenfalls, dass der digitale Workflow bisher nur bei der Gerütherstellung (CAM-Stufe) von implantatgetragenen Restaurationen gegenüber konventionell gegossenen Strukturen Genauigkeitsvorteile aufweist. Für die vorausgehenden Schritte, wie Abformung und Modellherstellung, konnte keine

Überlegenheit des digitalen Workflows gegenüber analogen Verfahren nachgewiesen werden [39].

3.3 Digitaler Workflow

3.3.1 Digitale Abformung

3.3.1.1 Intraoralscanner

IOS ermöglichen die direkte Erfassung von Oberflächenstrukturen durch das Zusammensetzen einer Vielzahl einzelner Bildaufnahmen zu einem vollständigen 3D-Datensatz. Der Datensatz wird als Punktwolke dargestellt, wobei jeder Punkt durch eine x-, y- und z-Koordinate definiert ist. Um eine Beurteilung der Oberfläche zu ermöglichen, wird die Punktwolke in der Scanner-Software als farbiges 3D-Modell visualisiert [40]. Dieses kann in standardisierten Formaten wie Standard Tessellation Language (STL), Polygon File Format (PLY) oder Objekt (OBJ) exportiert werden [19]. Während der digitalen Abformung wird der 3D-Datensatz durch die Verknüpfung mehrerer Einzelbilder generiert, wobei die Randbereiche der aufgenommenen Bilder überlagert werden. Ungenauigkeiten beim Überlagern einzelner Bilder können sich zu größeren Fehlern im Gesamtdatensatz aufsummieren [41]. Die Zahl dieser Verknüpfungsfehler steigt dabei mit der Anzahl der zu verknüpfenden Einzelbilder sowie durch die Erfassung von weniger strukturierten Bereichen [23, 27].

Häufig verwendete Aufnahmeprinzipien sind die konfokale Mikroskopie (z.B. 3Shape Trios 5, CEREC Primescan) und die optische Triangulation (z.B. CEREC Omnicam) [42, 43]. Bei der optischen Triangulation wird ein Lichtstrahl oder Streifenmuster auf die Zahnoberfläche projiziert. Aus der geometrischen Beziehung zwischen Projektor, Sensor und reflektiertem Punkt kann die räumliche Lage des Objektes berechnet werden [44]. Dieses Verfahren ermöglicht eine schnelle Datenerfassung. Es zeigt jedoch Einschränkungen bei spiegelnden Oberflächen durch die Ablenkung des reflektierten Lichtstrahls [45]. Die konfokale Mikroskopie basiert auf einem fokussierten Laserstrahl, dessen Reflexion durch eine Lochblende detektiert wird. Nur Licht aus der Fokusebene wird registriert, während Streulicht blockiert wird. Dadurch entstehen hochauflösende, kontrastreiche Schichtaufnahmen [45]. Durch die sequentielle Erfassung mehrerer Tiefenebenen liefert diese Technik eine hohe Tiefenselektion und Schärfe [35]. IOS, wie der Trios 5, arbeiten mit reiner konfokaler

Mikroskopie, während der CEREC Primescan konfokale Mikroskopie mit weiterentwickelter Projektions- und Sensortechnik ("Dynamic Depth Scan") kombiniert [42]. In der Literatur zeigt sich, dass konfokale Systeme sowie deren Hybridsysteme häufig triangulationsbasierten Systemen überlegen sind [26, 43, 46, 47].

3.3.1.2 Photogrammetrie

Photogrammetrie (PG)-Systeme nutzen hochauflösende Fotografien aus unterschiedlichen Perspektiven zur Rekonstruktion von 3D-Objekten. In der Zahnmedizin kommen hierfür Systeme zum Einsatz, die Implantatpositionen durch die Aufnahme mehrerer extraoraler Bilder erfassen [48]. Die Erfassung der Implantatpositionen erfolgt dabei durch das gleichzeitige Anvisieren von Markierungen auf speziellen Scanbodies (sog. Markern). Zahnmedizinische PG-Systeme basieren zwar auf dem Grundprinzip der PG, verfügen jedoch über spezialisierte Softwareprogramme, deren Algorithmen an die spezifischen implantatprothetischen Anforderungen angepasst sind [49]. Eine Erfassung des Weichgewebes ist jedoch nicht möglich [50], sodass in den meisten Fällen eine Kombination aus PG zur Erfassung der Implantatposition und IOS zur Weichgewebeaufnahme erforderlich ist. Die resultierenden Datensätze werden abschließend in der CAD-Software zusammengefügt. Im Vergleich zu konventionellen Abformmethoden und auch zu digitalen Abformungen mit IOS zeichnen sich PG-Abformungen insbesondere durch ihre hohe Genauigkeit, Geschwindigkeit und erhöhten Patientenkomfort aus [49]. Die Genauigkeit extraoraler PG-Systeme variiert in verschiedenen Studien zwischen 10 und 77 μm (Richtigkeit) sowie 2 und 203 μm (Präzision) [48]. Da sie nicht durch die typischen Limitationen (Verknüpfungsfehler, äußere Einflüsse) beeinträchtigt werden, stellen PG-Systeme eine vielversprechende Alternative zur digitalen Abformung mit IOS dar [51]. Einige Studien zeigen, dass PG-Abformungen bei implantatbasierten Ganzkieferabformungen eine höhere Richtigkeit und Präzision erreichen als konventionelle und digitale Abformungen mit IOS [52, 53]. In der Literatur finden sich jedoch kontroverse Ergebnisse hinsichtlich einer möglichen Überlegenheit von PG-Systemen [54, 55].

3.3.2 CAD

3.3.2.1 *Digitales Designen von Restaurationen*

Der CAD-basierte Entwurf von Restaurationen bildet die Schnittstelle zwischen der digitalen Erfassung der klinischen Situation und der nachfolgenden computer-gestützten Fertigung der Restaurationen [56]. In den vergangenen zwei Jahrzehnten wurden unterschiedliche Ansätze zum digitalen Designen von Restaurationen entwickelt. Diese beruhen entweder auf statistisch-morphologischen Modellen, der Anpassung standardisierter Zahnvorlagen aus Referenzbibliotheken oder der Spiegelung kontralateraler Zähne [57]. Ein wichtiger Meilenstein in der automatisierten Restaurationserstellung ist das von Mehl entwickelte statistische biogenerische Zahnmodell [58, 59]. Es basiert auf einer digitalen Referenzdatenbank, aus der ein Durchschnittszahn berechnet wird [60, 61]. Mittels Hauptkomponentenanalyse werden typische Formvariationen erfasst, sodass fehlende Strukturen durch eine Kombination aus Durchschnittszahn und gewichteten Formabweichungen rekonstruiert und patientenspezifisch angepasst werden [62]. Ein anderer Ansatz zur digitalen Zahnrekonstruktion beruht auf der Spiegelung kontralateraler Zähne [63]. Die auf diese Weise gewonnenen Zahnformen werden unter Berücksichtigung okklusaler Parameter digital ausgerichtet und optimiert. Weiterentwickelte Verfahren ermöglichen eine halbautomatisierte Segmentierung der Kiefer sowie eine Anpassung durch digitale Okklusionsalgorithmen [64].

Obwohl aktuelle CAD-Systeme über diese automatisierten Funktionen verfügen, bleibt die manuelle Nachbearbeitung durch den Zahntechniker unerlässlich. Er passt den Softwarevorschlag unter Berücksichtigung prothetischer Anforderungen, individueller Fallbesonderheiten und eigener Erfahrung an [57, 65]. Aktuelle Studien zeigen, dass KI, insbesondere Deep-Learning-Methoden, den bereits teilautomatisierten CAD-Prozess verbessern kann [65, 66]. Im Gegensatz zu klassischen Verfahren mit Referenzbibliotheken oder Spiegelungen erzeugen neuronale Netzwerke neue Zahnformen, indem sie charakteristische Muster und morphologische Strukturen selbstständig anhand der Morphologie der vorhandenen Restbezaehlung erlernen [57]. Von besonderer Bedeutung sind dabei dreidimensionale Generative Adversarial Networks (3D-GANs), die in der Lage sind, biometrische Zahnersatzdesigns weitgehend automatisiert zu erstellen [65]. Die Studienlage ist jedoch derzeit noch zu limitiert, um eindeutig zu beurteilen, ob KI-basierte CAD-Restaurationsdesigns einen

Vorteil hinsichtlich der Genauigkeit im Vergleich zu klassischen CAD-Designs bieten [66, 67].

3.3.2.2 Minimalinvasive Restaurationskonzepte im Seitenzahnbereich

Die minimalinvasive Zahnmedizin verfolgt das Ziel, möglichst viel gesunde Zahnhartsubstanz zu erhalten und nur so viel zu präparieren, wie für die Aufnahme einer funktional und ästhetisch belastbaren Restauration erforderlich ist [68]. Abhängig vom vorhandenen Zahnschaden lassen sich die folgenden Restaurationstypen zur Versorgung von Defekten im Seitenzahnbereich unterscheiden: Vollkronen als maximal umschließende Versorgung bei großflächigerem Substanzverlust [69], Teilkronen als Zwischenstufe zur minimalinvasiven Versorgung von Teildefekten mit zusätzlicher funktioneller Unterstützung [70] und okklusale Veneers als maximal substanzschonende, non-retentive Teilrestorationen [71]. Durch die zusätzliche Einbeziehung von Höckern und/oder approximalen Flächen ermöglichen Teilkronen eine verbesserte Stabilisierung des Restzahnes und eine gleichmäßigere Kraftübertragung, ohne den umfassenden Substanzabtrag einer Vollkrone zu erfordern. Während für Vollkronen eine Zahnschadenreduktion von etwa 67,5 % bis 75,6 % beschrieben wird, liegt der Substanzverlust bei minimalinvasiven Restaurationen, wie Inlays und Onlays mit etwa 20 % und 39 % deutlich niedriger [72]. In vitro Studien zeigen, dass auch non-retentive okklusale Veneers in stark reduzierten Schichtstärken (≤ 0.5 mm) belastbar sein können, sofern ein geeignetes Restaurationsmaterial und eine adäquate adhäsive Befestigung gewählt wird [73]. Sie stellen eine minimalinvasive Behandlungsoption für Patienten mit erosivem und/oder abrasivem Zahnhartsubstanzverlust dar, um die okklusale vertikale Dimension wiederherzustellen [71]. Klinische Studien, die minimalinvasive Restaurationen (< 1 mm) untersuchen, sind jedoch selten [74] und limitiert, da zahlreiche weitere Faktoren, wie das Adhäsivprotokoll, die Wahl des Keramikmaterials, das Präparationsdesign, Abform- und Herstellungsmethoden sowie die Erfahrung des Zahnarztes und Technikers den Erfolg der Restauration beeinflussen [75]. Auch das Ermüdungsverhalten minimalinvasiver CAD/CAM-Restaurationen unter zyklischer Kautbelastung sowie die physikalischen Versagensmechanismen keramischer Werkstoffe sind bislang nicht hinreichend untersucht.

3.3.3 CAM

3.3.3.1 Digitale Fertigungstechniken

Digitale Fertigungstechniken lassen sich grundlegend in subtraktive, additive und zunehmend hybride Technologien unterteilen [76, 77]. Bei subtraktiven Verfahren, wie computergestütztem Fräsen oder Schleifen, erfolgt ein Materialabtrag von vorkonfektionierten Rohlingen (z.B. aus Keramik, Komposit oder Metall) [78]. Gefräste Restaurationen zeichnen sich durch eine hohe Passgenauigkeit und Oberflächengüte aus, insbesondere wenn 5-Achs-Fräseinheiten mit einem geeigneten Fräsprotokoll verwendet werden [79]. Im Gegensatz zu herkömmlichen 3-Achs-Maschinen können so fünf Achsen gleichzeitig bewegt werden, was die Bearbeitung komplexer Geometrien und schwer zugänglicher Oberflächen erleichtert [80]. Beispiele in der restaurativen Zahnmedizin sind die Herstellung von Kronen- und Teilkronenrestaurationen, Brücken, Implantatabutments sowie Gerüststrukturen [81].

Additive Verfahren, vor allem verschiedene 3D-Druck-Technologien, ermöglichen dagegen einen schichtweisen Materialaufbau [82]. Sie erlauben die Herstellung komplexer, individualisierter Geometrien mit hoher Designfreiheit und geringem Materialverlust [8]. Einsatzgebiete sind derzeit u.a. die Herstellung von Bohrschablonen, provisorischen Kronen und Brücken, individualisierten Abformlöffeln sowie Okklusionsschienen. Darüber hinaus finden sie Anwendung in der Fertigung von Modellen, Interimsprothesen und Gerüststrukturen [83]. Obwohl 3D-gedruckte Materialien in der Zahnmedizin eine zunehmend höhere Präzision und Biokompatibilität aufweisen, zeigen Studien, dass diese Werkstoffe zytotoxische Effekte auf Gingiva-Keratinocyten hervorrufen können. Dabei scheinen insbesondere der Monomeranteil sowie die Zusammensetzung der Füllkörper eine entscheidende Rolle zu spielen [84, 85]. Unpolymerisierte Monomere können in die orale Umgebung gelangen, verschluckt oder über Dentintubuli in die Pulpa transportiert werden [86], wo sie Überempfindlichkeitsreaktionen sowie zytotoxische Effekte auf lokale Gewebszellen hervorrufen können [87]. Der Polymerisationsgrad von Kunststoffen wird maßgeblich durch Parameter des Verarbeitungsprozesses, wie die Dauer und Temperatur der Polymerisation sowie - bei Photopolymerisation - der Belichtungszeit und Intensität beeinflusst [88]. Die aktuelle Studienlage deutet darauf hin, dass durch

gezielte Modifikationen der Zusammensetzung und durch effektive Nachhärtungsprotokolle die Biokompatibilität 3D-gedruckter Restaurationen verbessert werden kann. Aktuelle in vitro Studien konnten zeigen, dass nachgehärtete 3D-gedruckte Komposite hinsichtlich der Monomerfreisetzung eine bessere Biokompatibilität als konventionelle direkte Komposite aufwiesen [89]. Zudem wurde nachgewiesen, dass die Zugabe von 5 % Glasfüllern sowie 10-20 % Zirkoniumoxid-Nanopartikeln die Biokompatibilität von 3D-gedrucktem PMMA signifikant verbessern kann [86]. Diese Erkenntnisse lassen die langfristige klinische Anwendung 3D-gedruckter Materialien zunehmend vielversprechend erscheinen.

3.3.3.2 CAM-geeignete Materialien

CAM-geeignete restaurative Werkstoffe lassen sich grundsätzlich in vier Hauptgruppen unterteilen: Keramiken, Hybridkeramiken, Polymere und Metalle [90, 91]. Unter den keramischen Werkstoffen zählen insbesondere Silikatkeramiken, wie feldspatbasierte Keramiken und Glaskeramiken (z.B. Lithiumdisilikat [LDS]) sowie Zirkoniumoxidkeramiken, zu den derzeit klinisch relevantesten Materialien in der restaurativen Zahnmedizin [92]. Sie zeichnen sich durch eine hohe Biokompatibilität, Farb- und Formstabilität sowie eine ausgezeichnete Ästhetik aus [93]. Hybridmaterialien wie CAD/CAM-Komposite oder polymerinfiltrierte Keramiknetzwerke (PICN) vereinen keramische und polymerbasierte Strukturen und zeichnen sich durch eine erhöhte Elastizität sowie eine gute Reparaturfreundlichkeit aus [94, 95]. Polymere Werkstoffe, insbesondere PMMA-basierte Materialien, werden vor allem für die Herstellung von Provisorien, individuellen Löffeln, Bisschablonen sowie 3D-gedruckten Modellen verwendet [90]. Metalle, insbesondere CoCr-Legierungen und Titan, finden vorrangig Anwendung in Gerüststrukturen, Stegen und Implantatabutments [91, 96]. Im Rahmen dieser Habilitationsschrift liegt der Schwerpunkt auf keramischen Werkstoffen, insbesondere LDS und verschiedenen Zirkoniumoxid-Generationen, die für definitive Restaurationen wie Kronen, Teilkronen und okklusale Veneers von besonderer klinischer Relevanz sind. Zudem wurde Titan als Material für die CAD/CAM-Fertigung implantatgetragener Stege berücksichtigt.

LDS-Glaskeramiken haben sich aufgrund ihrer günstigen Kombination aus Ästhetik und klinischer Langzeitzuverlässigkeit als eines der zentralen Keramikmaterialien im

CAD/CAM-Workflow etabliert [97]. Sie zeichnen sich durch eine hohe Biegefestigkeit (ca. 400-600 MPa), einem dem Schmelz vergleichbaren Elastizitätsmodul (LDS: ~95 GPa, Schmelz: ~70 GPa), einem verschleißfreundlichen Antagonistenverhalten sowie einem zuverlässigen adhäsiven Verbund aus [98]. Daher eignen sie sich besonders gut für minimalinvasive Einzelzahnrestaurationen im Front- und Seitenzahnbereich. Zudem sind sie gemäß der Herstellerangaben für dreigliedrige Brücken einschließlich des zweiten Prämolaren zugelassen [99]. Langzeituntersuchungen belegen eine hohe Überlebenswahrscheinlichkeit: Für CAD/CAM-gefertigte LDS-Kronen wurden Überlebensraten von über 90 % nach sechs Jahren [100] sowie über 80 % nach 15 Jahren [101] beschrieben. Ein wesentlicher Vorteil des Materials liegt in der Möglichkeit, auch dünnwandige Versorgungen wie okklusale Veneers herzustellen [102]. Während Hersteller für Seitenzahnkronen Mindestschichtstärken von 1 mm empfehlen, zeigen in vitro und erste klinische Studien, dass minimalinvasive Schichtstärken zwischen 0,75-1,0 mm möglich sind [71, 74].

Zirkoniumoxidkeramiken haben sich parallel zu LDS von einem klassischen Gerüstmaterial zu einem vielseitigen Restaurationswerkstoff weiterentwickelt [103]. Konventionelles 3Y-TZP (3 mol% yttriumoxidstabilisiertes tetragonales Zirkoniumoxid-Polykristall) überzeugt durch eine sehr hohe Festigkeit (1000–1400 MPa), weist jedoch nur eine begrenzte Transluzenz auf [93]. Um diese ästhetische Limitation zu überwinden, wurden Varianten mit erhöhtem Yttriumgehalt entwickelt (4Y-PSZ, 5Y-PSZ) [104]. Der damit verbundene Anstieg des kubischen Phasenanteils steigert zwar die Transluzenz, geht aber mit einer reduzierten mechanischen Belastbarkeit einher (ca. 400–1000 MPa je nach Generation) [93]. Damit rücken die mechanischen Eigenschaften transluzenter Zirkoniumoxide in den Bereich von LDS, was sie für minimalinvasive Indikationen wie Veneers oder Teilkronen interessant macht, auch wenn solche Indikationen bislang nicht in den offiziellen Herstellerempfehlungen enthalten sind [105].

Eine wesentliche Innovation stellen mehrschichtige Zirkoniumoxide dar. Während frühe Varianten lediglich einen Farbverlauf bei homogener Materialzusammensetzung boten, kombinieren aktuelle Systeme unterschiedliche Materialgenerationen in einem Rohling [104]. Dabei wurden hochfeste, überwiegend

tetragonale Bereiche im zervikalen Anteil mit transluzenten, kubischen Strukturen in den inzisalen oder okklusalen Zonen kombiniert [106]. So ließen sich Festigkeit und Ästhetik innerhalb einer Restauration vereinen. Allerdings beeinflussen diese Materialgradienten auch das Bruchverhalten. Kritische Defekte können je nach Positionierung der Restauration im Rohling in unterschiedlichen Schichten entstehen [107]. Vor allem in der Übergangszone zwischen den transluzenteren und hochfesten Schichten ist das Materialverhalten bislang nur unzureichend untersucht und der klinische Erfolg der Restauration unklar.

Titan wird in der restaurativen Zahnmedizin vor allem für die Herstellung von Gerüsten, Stegen, individuellen Abutments und implantatgetragenen Zahnersatz verwendet [108]. Durch seine hervorragende Biokompatibilität, hohe Festigkeit und Korrosionsbeständigkeit gilt Titan als bevorzugtes Material für anspruchsvolle prothetische Versorgungen, insbesondere im Bereich der Implantatprothetik [109]. Im Vergleich zu herkömmlich gegossenem Titan weisen CAD/CAM-gefräste Titanstrukturen signifikante Vorteile in Bezug auf ihre Oberflächenmikrostruktur und biomechanischen Eigenschaften auf [91, 110]. Das Fräsen aus homogenisierten Titanblöcken ermöglicht sehr glatte und dichte Oberflächen, wodurch das Risiko von Lunkern, Porositäten und Mikrorissen, wie sie beim Guss auftreten können, deutlich minimiert wird. Diese homogene Mikrostruktur trägt maßgeblich zu einer verbesserten Passgenauigkeit und Spannungsfreiheit bei, was insbesondere bei großspannigen Stegkonstruktionen essentiell ist [111]. Klinische Langzeitdaten von Prothesen mit CAD/CAM-gefertigten Titanstegen und individuellen Abutments zeigen sehr gute Überlebensraten und eine hohe Gewebestabilität [112]. Neben dem subtraktiven Fräsen gewinnen additive Fertigungsverfahren wie das selektive Laserschmelzen (SLM) für die Herstellung von Titanstrukturen zunehmend an Bedeutung [113]. Diese hybriden oder rein additiven Verfahren erlauben eine noch größere Gestaltungsfreiheit und somit die Fertigung besonders komplexer Formen [77]. Aktuelle Forschungsergebnisse bestätigen, dass additiv gefertigte Titangerüste klinisch akzeptable Diskrepanzen aufweisen [114]. Langzeitdaten zur klinischen Anwendung additiv gefertigter Titanprothetik fehlen jedoch bisher [115].

4. Fragestellungen

1.) Wie wirkt sich die Wahl unterschiedlicher digitaler Abformverfahren (PG vs. IOS) auf die Genauigkeit implantatgetragener Ganzkieferrekonstruktionen aus?

Digitale Abformungen erreichen bei Einzelzahnversorgungen eine mit konventionellen Verfahren vergleichbare Genauigkeit. Sie stoßen jedoch bei Ganzkieferrekonstruktionen, insbesondere in zahnlosen Kiefern, an ihre Grenzen. Während IOS durch Scanpfade, Umgebungsbedingungen und Anwendererfahrungen beeinflusst werden, zeigen erste Studien, dass photogrammetrische Verfahren durch die extraorale Erfassung der Implantatposition kumulative Abweichungen verringern können. Eine eindeutige Überlegenheit der PG gegenüber IOS konnte angesichts der uneinheitlichen Evidenzlage bislang nicht nachgewiesen werden. Die in der Literatur berichteten Ergebnisse beziehen sich zudem ausschließlich auf die Abformungsebene. Ungeklärt bleibt insbesondere, in welchem Ausmaß sich die Unterschiede zwischen den Verfahren auf die Genauigkeit des gesamten digitalen Workflows auswirken. Die erste eingeschlossene Studie dieser Arbeit untersucht daher die vertikale Passung CAD/CAM-gefertigter, implantatgetragener Ganzkiefer-Titan-Gerüste, basierend auf PG- und IOS-Abformungen.

2.) Inwieweit beeinflussen Materialauswahl, Restaurationsdesign und reduzierte Schichtstärke die mechanische Leistungsfähigkeit keramischer Versorgungen innerhalb des digitalen Workflows?

Neben der Abformgenauigkeit bestimmt auch die Wahl des Restaurationsmaterials maßgeblich den klinischen Erfolg digitaler Workflows. Durch CAD/CAM-Technologien ist heute eine hochpräzise Bearbeitung bruchfester Keramikmaterialien in minimalinvasiven Schichtstärken (<1 mm) möglich. Allerdings liegen bislang nur begrenzte Daten zum Ermüdungsverhalten dünnwandiger keramischer Restaurationen, insbesondere mehrschichtiger Zirkoniumoxidvarianten, und deren zugrunde liegenden physikalischen Versagensmechanismen vor. Zudem fehlen Studien zum Vergleich verschiedener Restaurationsdesigns. Ein vertieftes Verständnis dieser Prozesse ist erforderlich, um das Materialverhalten und dessen Einflussfaktoren zuverlässig einschätzen und evidenzbasierte Empfehlungen zur Schichtdickenreduktion für den klinischen Einsatz ableiten zu können. Drei Studien dieser Arbeit untersuchen daher den Einfluss des Restaurationsmaterials, -designs

und der Schichtstärke auf die Bruchfestigkeit CAD/CAM-gefertigter Einzelzahnrestorationen (Kronen, Teilkronen und okklusale Veneers). In einer weiteren Studie erfolgte eine fraktographische Analyse zur Charakterisierung der Rissausbreitung. Ergänzend wurde ein mathematisches Modell angewandt, um die theoretische Lebensdauer der Restorationen abzuschätzen und mit den experimentellen Ergebnissen zu vergleichen.

5. Originalarbeiten

5.1 Originalarbeit 1

In vitro comparison between complete-arch implant supported frameworks using intraoral scanning and photogrammetry.

5.1.1 Hintergrund

Im Rahmen der Digitalisierung etablieren sich sowohl IOS als auch photogrammetrische Systeme für die Herstellung von implantatgetragenen Ganzkiefer-Restaurationen zunehmend in der klinischen Praxis [116]. Beide Technologien versprechen eine Verbesserung der Arbeitsabläufe, einen erhöhten Patientenkomfort sowie eine Reduktion möglicher Fehlerquellen im Vergleich zur klassischen konventionellen Abformung [51, 55]. Jedoch variieren die berichteten Ergebnisse hinsichtlich ihrer Genauigkeit [54, 117]. Darüber hinaus beschränkt sich die derzeitige Evidenz zur Anwendung photogrammetrischer Systeme im Gesamtworkflow bislang auf Fallberichte [118, 119]. Ziel der vorliegenden Studie war es deshalb, die vertikale Passung implantatgetragener Ganzkiefer-Titan-Gerüste hergestellt auf der Grundlage von IOS und PG-Abformungen zu untersuchen und miteinander zu vergleichen.

5.1.2 Material und Methode

In dieser Studie wurde ein zahnloses Unterkiefermodell (Knochenqualität Typ 2) verwendet. Es wurden vier Implantate in den Positionen 34, 32, 42 und 44 (nach dem Fédération Dentaire Internationale [FDI] System) inseriert und mit Multiunit-Abutments sowie systemspezifischen Abformpfosten (Scanbodies) versehen. Um eine Referenzgruppe zu erzeugen, wurde das Modell zunächst mit einem hochgenauen Laborscanner (InEosX5, Genauigkeit: $2,1 \pm 2,8 \mu\text{m}$) digitalisiert. Die folgenden Testscanner wurden untersucht: IOS (Primescan) sowie zwei PG-Systeme (ICam4D und PIC). Für jede Testgruppe wurden je zwei digitale Ganzkiefer-Abformungen erstellt, auf denen in einem autorisierten Fräs-Center CAD/CAM gefertigte Titangerüste hergestellt wurden. Die Ermittlung der Passgenauigkeit erfolgte über die quantitative Analyse der vertikalen Mikrospalten (Microgaps) mittels modifiziertem Sheffield-Test. Dafür wurden die Titangerüste zunächst mit allen

Schrauben auf den Multiunit-Abutments des Referenzmodells fixiert (5N). Die Schrauben wurden dann wieder entfernt mit Ausnahme einer Schraube in Position 32. Im Anschluss wurden detaillierte Messungen des Abstands zwischen dem jeweiligen Abutment (34, 42 und 44) und dem gelösten Titangerüst mithilfe eines Rasterelektronenmikroskops (REM) bei 1600-facher Vergrößerung durchgeführt. Insgesamt wurden 27 Messungen (9 Messungen x 3 Implantate) pro Titangerüst vorgenommen. Die Datenanalyse erfolgte deskriptiv ($\text{mean} \pm \text{SD}$) sowie mittels ANOVA und anschließendem Tukey's post hoc Test ($p=0,05$).

5.1.3 Ergebnisse

Die geringsten vertikalen Mikrospalt-Werte ($22,65 \pm 7,28 \mu\text{m}$) wurden in der Referenzgruppe (Laborscanner) gemessen. Die Ergebnisse der PG-Systeme lagen nahe an der Referenz (ICam: $24,1 \pm 7,99 \mu\text{m}$, PIC: $30,52 \pm 9,81 \mu\text{m}$). Die durchschnittlichen Werte der IOS-Abformung ($33,21 \pm 13,84 \mu\text{m}$) waren verglichen mit dem Laborscanner ($p=0,017$) und der ICam ($p=0,049$) signifikant höher. Es bestanden keine statistisch signifikanten Unterschiede zwischen beiden PG-Systemen ($p=0,344$). Die größten vertikalen Mikrospalt-Werte traten bei der Implantatposition 44 auf, vor allem in den Gruppen PS und PIC. In Position 34 zeigten sich keine signifikanten Unterschiede zwischen den Gruppen. In Position 42 bestanden signifikante Unterschiede zwischen Referenz und PS ($p<0,001$), PS und ICam ($p<0,001$), PS und PIC ($p<0,001$) sowie ICam und PIC ($p=0,005$). In Position 44 wurden ebenfalls zwischen Referenz und PS ($p<0,001$), PS und ICam ($p<0,001$), Referenz und PIC ($p<0,001$) sowie ICam und PIC ($p<0,001$) signifikante Unterschiede festgestellt.

5.1.4 Diskussion

Die Ergebnisse dieser Studie zeigen, dass implantatgetragene Ganzkiefer-Restaurationen basierend auf PG-Abformungen hinsichtlich ihrer Passung den Restaurationen, die mit einem IOS erzeugt wurden, überlegen waren. IOS erzeugen aus Einzelbildern einen 3D-Gesamt-Datensatz, wobei Überlagerungsfehler insbesondere beim Scan größerer Areale oder wenig strukturierter zahnloser Kiefer auftreten können [120, 121]. Zahnmedizinische PG-Systeme nutzen dagegen Referenzpunkte auf speziellen Scanbodies und extraorale Kameras, wodurch patientenbedingte Störfaktoren weitgehend reduziert werden [122]. Die dabei

entstehenden Fehler können während des digitalen Design- und Fräsprozesses kumulieren und zu vertikalen Diskrepanzen führen, die den passiven Sitz der Restauration gefährden können [54]. Die in dieser Studie gewonnenen Ergebnisse bestätigen frühere Forschungsergebnisse, welche PG als überlegende Technik zur Erfassung implantatgetragener zahnloser Kiefer beschreiben [117, 123, 124]. In der vorliegenden Studie erreichten beide photogrammetrischen Systeme (ICam und PIC) ähnliche Werte bezüglich des vertikalen Mikrospalts, was darauf hindeutet, dass die Technik der PG konstante und herstellerunabhängige Ergebnisse liefert. Die signifikante Zunahme des vertikalen Mikrospalts bei entfernteren Implantaten (Position 44) verdeutlicht zudem, dass längere Strecken oder ungünstige Positionen von Implantaten auch für dieses Verfahren eine Herausforderung darstellen. Alle gemessenen Diskrepanzen lagen jedoch deutlich innerhalb des am häufigsten in der Literatur als klinisch akzeptabel beschriebenen Grenzwerts von 150 μm [39]. Dies spricht dafür, dass beide Technologien (IOS und PG) klinisch einsetzbar sind, wobei PG-Systeme aufgrund der geringeren Fehleranfälligkeit für die Erfassung von zahnlosen Kiefern mit Implantaten präferiert werden könnten. Zukünftige Studien sollten neben der klinischen Validierung der Ergebnisse auch die langfristigen Auswirkungen der beobachteten Mikrospalt-Unterschiede auf die Langlebigkeit und Patientenzufriedenheit untersuchen.

5.2 Originalarbeit 2

Monolithic zirconia crowns: Effect of thickness reduction on fatigue behavior and failure load.

5.2.1 Hintergrund

Monolithische Zirkoniumoxidkronen haben sich in der restaurativen Zahnmedizin aufgrund ihrer hohen Bruchfestigkeit und reduzierten Anfälligkeit für technische Komplikationen, wie Chipping, etabliert [125]. Die hohe Bruchfestigkeit (3Y-TZP: 1000-1400 MPa) macht sie attraktiv für Restaurationen mit reduzierter Materialstärke und ermöglicht substanzschonende Präparationstechniken [93]. Obwohl monolithische 3Y-TZP Zirkoniumoxidkronen im Vergleich zu verblendeten und LDS-basierten Kronen höhere Bruchlasten erreichen, scheinen sie in reduzierten Schichtstärken eine erhöhte Anfälligkeit für Komplettfrakturen aufzuweisen [126, 127].

Im funktionell höher beanspruchten Seitenzahnbereich ist bislang nicht eindeutig geklärt, welche Mindeststärke unter klinisch relevanten Belastungsbedingungen eine ausreichende Stabilität gewährleistet. Ziel dieser in vitro Studie war es daher, den Einfluss unterschiedlicher okklusaler Schichtdicken auf die Bruchlast und das Ermüdungsverhalten monolithischer 3Y-TZP Zirkoniumoxidkronen im Molarenbereich unter standardisierten Testbedingungen zu untersuchen.

5.2.2 Material und Methode

In der vorliegenden Studie wurden 140 CAD/CAM-gefertigte monolithische Kronen aus 3Y-TZP-Zirkoniumoxid (inCorisTZI) mit fünf verschiedenen okklusalen Schichtstärken (2,0 mm, 1,5 mm, 1,0 mm, 0,8 mm, 0,5 mm; jeweils n=28) auf standardisierten Komposit-Stümpfen eines unteren ersten Molaren befestigt. Nach der Konditionierung mit einem MDP-haltigen Primer wurden die Kronen mit einem selbstadhäsiven Kompositzement eingesetzt. Zur Simulation einer fünfjährigen klinischen Belastung wurde die Hälfte der Kronen pro Gruppe (n=14) einem thermomechanischem Ermüdungstest in einem Kausimulator unterzogen (200 N, 1,6 Hz, 1,2 Millionen Zyklen, Temperaturwechsel 5-55 °C). Im Anschluss erfolgte bei allen Proben, sowohl bei belasteten als auch unbelasteten, eine Einzelbelastung bis zur Fraktur mittels Universalprüfmaschine zur Bestimmung der maximalen Bruchlast. Zur Frakturanalyse wurde ein Stereolichtmikroskop eingesetzt. Die statistische Auswertung erfolgte auf Basis einer Zwei-Parameter-Weibull-Verteilung zur Ermittlung der charakteristischen Festigkeit und der Überlebenswahrscheinlichkeit der jeweiligen Gruppen.

5.2.3 Ergebnisse

Alle Kronen mit einer okklusalen Schichtstärke von $\geq 1,0$ mm überstanden die Ermüdungsbelastung ohne sichtbare Schäden. Dagegen zeigten die Gruppen mit 0,8 bzw. 0,5 mm Schichtstärken nach der Kausimulation in 21 % bzw. 36 % der Fälle Rissbildungen. Die höchsten Bruchlastwerte wurden in der 2,0-mm-Gruppe mit einer charakteristischen Festigkeit von 5934 N (unbelastet) und 5540 N (belastet) gemessen, gefolgt von der 1,5-mm-Gruppe mit 5008 N (unbelastet) und 4576 N (belastet) sowie der 1,0-mm-Gruppe mit 2783 N (unbelastet) und 2785 N (belastet). Deutlich niedrigere Werte wurden bei Kronen mit reduzierten Schichtstärken (0,8 und

0,5 mm) beobachtet: Die 0,8-mm-Gruppe erreichte Werte von 1555 N (unbelastet) und 1533 N (belastet). Die 0,5-mm-Gruppe zeigte nur Bruchlastwerte von 1455 N (unbelastet), wobei letzterer nach Ermüdung auf 1122 N sank. Die Weibull-Analyse zeigte eine signifikant geringere Zuverlässigkeit bei den in der Schichtstärke reduzierten Gruppen, insbesondere bei 0,8 mm Schichtstärke mit einem Weibull-Modul von nur 2,89. Insgesamt hatte die Ermüdungsbelastung keinen signifikanten Einfluss auf die Bruchlast, mit Ausnahme der 0,5-mm-Gruppe. Die Frakturmodi waren in allen Gruppen vergleichbar und wiesen meist katastrophale Frakturen mit mehreren Fragmenten auf. Die Bruchlinien verliefen dabei typischerweise von der okklusalen Belastungsfläche zum zervikalen Kronenrand.

5.2.4 Diskussion

Die Ergebnisse der vorliegenden Studie verdeutlichen den maßgeblichen Einfluss der okklusalen Schichtstärke auf die mechanische Belastbarkeit monolithischer Zirkoniumoxidkronen aus 3Y-TZP. Es konnte gezeigt werden, dass eine Mindestschichtstärke von 1,0 mm ausreicht, um der simulierten fünfjährigen thermomechanischen Belastung standzuhalten. Demgegenüber zeigten jedoch Kronen mit Schichtstärken unter 1 mm, insbesondere mit 0,5 mm, eine signifikant verringerte Festigkeit nach fünfjähriger Ermüdungsbelastung. Diese Unterschiede wurden sowohl durch die gemessenen Bruchlastwerte als auch durch die statistische Analyse mittels Weibull-Verteilung bestätigt. Insgesamt hielten jedoch alle getesteten Kronen - unabhängig von der Wandstärke - Versagenslasten stand, die deutlich oberhalb der physiologischen Kaubelastungen im Molarenbereich lagen (50-250 N physiologisch, 500-900 N parafunktional [128]).

Bedeutsam ist die Feststellung, dass bereits geringe Unterschiede in der Wandstärke – etwa zwischen 0,8 mm und 1,0 mm – zu signifikanten Unterschieden in der mechanischen Belastbarkeit führen können. Dies betont die Relevanz einer präzisen Präparationskontrolle sowie einer sorgfältigen diagnostischen und präprothetischen Planung, insbesondere in klinischen Situationen mit eingeschränktem Platzangebot. Eine Unterschreitung der Schichtstärke von 1,0 mm scheint mit einem erhöhten Risiko für Rissbildung, Materialermüdung und vorzeitigem Versagen verbunden zu sein und sollte daher kritisch abgewogen werden. Entsprechend sprechen Hersteller die Empfehlung aus, eine okklusale Mindestschichtstärke von 1,0 mm für monolithische

3Y-TZP Zirkoniumoxidkronen nicht zu unterschreiten [105, 129]. Die gemessenen Bruchlastwerte dieser Studie sind vergleichbar mit anderen Studien, die 3Y-TZP Kronen in verschiedenen Schichtstärken untersuchten [127, 130]. Da jedoch Materialien anderer Hersteller und ein anderes Studiendesign verwendet wurden, ist ein direkter Vergleich nicht möglich.

Die häufig beobachteten katastrophalen Frakturen mit mehreren Fragmenten sowie die Position der Bruchlinien – typischerweise von der okklusalen Belastungszone zum zervikalen Rand – lassen typische Versagensmechanismen unter funktioneller Belastung erkennen. Auffällig war, dass bei den Gruppen mit 0,5 und 0,8 mm Schichtstärke sämtliche Kompositstümpfe unversehrt blieben. Dies kann auf ein verändertes Kraftverteilungsmuster zurückzuführen sein: Die geringere Kronenstärke führte zu einer früheren Materialüberbelastung im Zirkoniumoxid, sodass die Energie dissipiert wurde, bevor sie auf den Stumpf übertragen wurde. In den Gruppen mit $\geq 1,0$ mm Wandstärke war dagegen die höhere Steifigkeit der Kronen und die damit verbundene höhere Kraftüberleitung mit vermehrten Stumpffrakturen verbunden. Perspektivisch sollten zukünftige Studien insbesondere die neuen Generationen transluzenter Zirkoniumoxide (4Y- und 5Y-PSZ) untersuchen, da diese aufgrund veränderter Materialstruktur andere mechanische Eigenschaften aufweisen.

5.3 Originalarbeit 3

Minimally invasive CAD/CAM lithium disilicate partial-coverage restorations show superior in vitro fatigue performance than single crowns.

5.3.1 Hintergrund

Im Zuge des digitalen Wandels in der Zahnmedizin haben sich CAD/CAM-gefertigte Materialien, insbesondere LDS, als vielversprechende Option für substanzschonende Einzelzahnrestorationen im Seitenzahnbereich etabliert [131]. Für Zähne mit okklusalen und bukkal-zervikalen Defekten stellen diese Teilkronen (sog. „Full-Veneers“) besonders für Patienten mit starker Sichtbarkeit der Seitenzähne eine ästhetische und minimalinvasive Alternative zu konventionellen Teilkronen dar [132, 133]. Zahlreiche Studien belegen die guten mechanischen Eigenschaften sowie die zuverlässige adhäsive Befestigung von LDS [134-136]. Die Literatur zur minimal

erforderlichen Schichtstärke solcher Restaurationen ist jedoch kontrovers [71, 102]. Zudem gibt es keine Daten für minimalinvasive, non-retentive Teilkronen, die okklusal und bucco-zervikale Defekte fassen. Ziel der vorliegenden in vitro Studie war es daher, das Ermüdungsverhalten und die Bruchlast von CAD/CAM-gefertigten LDS-Teilkronen (Full-Veneers) mit unterschiedlichen okklusalen Schichtstärken (1,5 mm, 1,0 mm, 0,5 mm) im Vergleich zu Vollkronen nach thermodynamischer Belastung zu untersuchen.

5.3.2 Material und Methode

Insgesamt wurden 72 monolithische, CAD/CAM-gefertigte Restaurationen aus LDS-Keramik (IPS e.max CAD) untersucht. Es wurden drei verschiedene Schichtstärken (1,5 mm, 1,0 mm und 0,5 mm) und zwei Restaurationsdesigns (Teilkronen und Vollkronen) analysiert, woraus sich sechs Gruppen mit jeweils $n=12$ ergaben. Die Präparation eines Molaren erfolgte in der jeweiligen Schichtstärke an einem standardisierten Oberkiefer-Modell. Das Design der Teilkronen war non-retentiv und umfasste die okklusalen, bukkalen und proximalen Flächen. Zusätzlich wurden okklusal zwei diagonale Noppen zur besseren Positionierung präpariert. Die Kronenpräparation erfolgte analog, jedoch ohne okklusale Noppen. Nach digitaler Konstruktion (CAD) und Fertigung in einer 5-Achs-Fräseinheit (CAM), wurden alle Restaurationen kristallisiert, glasiert und adhäsiv mit einem dualhärtenden Befestigungszement auf den Kompositstümpfen befestigt. Diese wurden mit Hilfe von Silikonformen aus den präparierten Masterstümpfen hergestellt. Anschließend wurden alle Proben einer thermodynamischen Ermüdungsprüfung (49 N, 1,6 Hz, 1,2 Millionen Zyklen, Temperaturwechsel 5-55 °C) unterzogen. Unbeschädigte Proben zählten als „Erfolg“, während Proben mit Rissen oder Debondings, die jedoch weiterhin funktionsfähig waren, als „Überleben“ bewertet wurden. Danach erfolgte eine Einzelbelastung bis zum Bruch in einer Universalprüfmaschine. Die Frakturanalyse wurde lichtmikroskopisch sowie selektiv mittels REM durchgeführt. Die statistische Auswertung erfolgte mittels ANOVA und post-hoc-Test ($p<0,05$).

5.3.3 Ergebnisse

Die Gesamt-Überlebensrate war 100 % für alle LDS-Restaurationen. Es wurde kein Debonding sowie keine Keramikfraktur beobachtet. Jedoch zeigten acht Kronen und

eine Teilkrone - jeweils mit einer Schichtstärke von 0,5 mm – Rissbildungen nach simulierter Kaubelastung. Die Gesamt-Erfolgsrate betrug 87,5 % (Teilkronen: 96,9 % und Kronen: 75 %). Im anschließenden Einzelbelastungstest wiesen alle Teilkronen höhere Bruchlastwerte auf als die jeweiligen Kronen gleicher Schichtstärke. Bei 1,5 mm dicken Teilkronen lag die mittlere Bruchlast bei 2883 N gegenüber 2566 N bei den Kronen. Für 1,0 mm wurden 2451 N (Teilkronen) und 1798 N (Kronen) gemessen. In der Gruppe mit 0,5 mm Schichtstärke betrugen die Werte 2023 N (Teilkronen) und 1193 N (Kronen). Die Unterschiede waren bei 0,5 mm ($p < 0,001$) und 1,0 mm ($p = 0,006$) statistisch signifikant zugunsten der Teilkronen. Innerhalb der Kronengruppen zeigten reduzierte Schichtstärken signifikant geringere Bruchlasten. Bei den Teilkronen hingegen gab es zwischen 0,5 mm und 1,0 mm keine signifikanten Unterschiede. Die Frakturanalyse ergab, dass bei Restaurationen mit reduzierten Schichtstärken vermehrt kohäsive Frakturen ohne Beteiligung des Kompositstumpfes auftraten, wohingegen bei dickeren Restaurationen häufiger Frakturen mit Beteiligung des Stumpfes beobachtet wurden.

5.3.4 Diskussion

Die vorliegende Studie zeigt, dass sowohl das Restaurationsdesign als auch die keramische Schichtstärke einen signifikanten Einfluss auf die mechanische Stabilität von LDS-Restaurationen haben. Minimalinvasive Teilkronen zeigten bei identischer Schichtstärke durchgängig höhere Bruchlastwerte und wiesen eine höhere Erfolgsrate nach thermodynamischer Belastung auf als Vollkronen. Auffällig war das deutlich bessere Ermüdungsverhalten der ultradünnen (0,5 mm) Teilkronen, die nur in einem Fall Rissbildungen zeigten. Im Gegensatz dazu traten bei den Kronen mit der Schichtstärke 0,5 mm acht Rissereignisse auf. Zudem blieb bei Restaurationen mit reduzierten Schichtstärken der Stumpf häufiger unbeschädigt, was auf eine gewisse Schutzwirkung der Restauration hindeutet. Aus restaurativer Sicht ist dies von Vorteil, da die erhaltende Zahnschubstanz eine erneute Versorgung ermöglicht. Dagegen kann eine Fraktur des Stumpfes, insbesondere bei stark kompromittierter Restzahnschubstanz, im ungünstigen Fall zur Nichterhaltungswürdigkeit des Zahnes und einer notwendigen Extraktion führen.

Die beobachteten Unterschiede zwischen Teilkronen und Kronen lassen sich sowohl durch die bessere Widerstandsfähigkeit des Zahn-Restaurations-Komplexes infolge des Zahnsubstanzerhalts als auch durch die günstigere Geometrie und resultierende Spannungsverteilung erklären [72, 132]. Bei Vollkronen kann die vollständige Umfassung des Zahnes zu höheren Spannungen in den lateralen Wandbereichen führen, was das Risiko für Rissbildungen und Frakturen begünstigt [137, 138]. Die weniger umgreifende Gestaltung der Teilkronen scheint hingegen eine gleichmäßigere Kraftverteilung zu ermöglichen. Der Einfluss der Schichtstärke zeigte sich insbesondere bei den Kronen signifikant, wohingegen bei den Teilkronen zwischen 0,5 mm und 1,0 mm keine signifikanten Unterschiede festgestellt wurden. Dies stellt einen Hinweis auf die mechanische Stabilität des Teilkronendesigns auch bei reduzierten Schichtstärken dar.

In der aktuellen Literatur werden vergleichbare Überlebensraten für Teilkronen und Vollkronen berichtet [135, 139]. Allerdings untersuchte keine dieser Studien Restaurationen mit minimalinvasiven Schichtstärken. Bei reduzierter Schichtstärke kann das Restaurationsdesign an Bedeutung gewinnen [74]. Für die klinische Anwendung deuten die Ergebnisse dieser in vitro Studie darauf hin, dass minimalinvasive Teilkronen aus LDS insbesondere in Situationen mit begrenztem okklusalem Platzangebot oder bei kombiniert okklusal-zervikalen Defekten eine substanzschonende und gleichzeitig mechanisch belastbare Alternative zu Vollkronen darstellen können. Zukünftige Studien sollten klären, inwiefern diese Ergebnisse auf andere Keramiken, insbesondere transluzente (4Y- und 5Y-PSZ) sowie mehrschichtige Zirkoniumoxide, übertragbar sind.

5.4 Originalarbeit 4

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

5.4.1 Hintergrund

Zahnhartsubstanzverlust bedingt durch Erosion hat sich in den vergangenen Jahrzehnten zu einem bedeutsamen oralen Gesundheitsproblem mit einer weltweit steigenden Prävalenz entwickelt, die bei Erwachsenen je nach Studie zwischen 2 %

und nahezu 100 % variiert [140]. Okklusale Veneers - Onlays mit minimalinvasiver Schichtstärke und non-retentivem Präparationsdesign [71] – stellen hierbei eine substanzschonende Alternative zu konventionellen Vollkronen dar, insbesondere bei jungen Patienten [102]. Die Einführung mehrschichtiger Zirkoniumoxide, bestehend aus einem festeren, basal/zervikalen Anteil (3Y-TZP/4Y-PSZ) und einem transluzenteren, inzisal/okklusalen Anteil (5Y-PSZ), eröffnet neue Möglichkeiten für hochästhetische und gleichzeitig mechanisch belastbare Restaurationen [93, 103]. Die mechanischen Eigenschaften der Übergangszone zwischen diesen Schichten sind bislang jedoch unzureichend erforscht, da ihre experimentelle Bestimmung technisch sehr anspruchsvoll ist. Ziel der vorliegenden in vitro Studie war es daher, das Ermüdungsverhalten und die Bruchlast von okklusalen Veneers aus kompositionsgradientem, mehrschichtigem Zirkoniumoxid (Übergangszone zwischen 4Y- und 5Y-PSZ) sowie aus LDS in drei unterschiedlichen Schichtstärken (0,5 mm, 1,0 mm und 1,5 mm) zu untersuchen.

5.4.2 Material und Methode

Untersucht wurden 72 CAD/CAM gefertigte okklusale Veneers (n=12 je Gruppe) aus LDS (IPS e.max CAD) und kompositionsgradientem, mehrschichtigem Zirkoniumoxid (IPS e.max ZirCAD Prime Esthetic). An einem Molaren eines Oberkiefer-Typodontmodells wurden zunächst drei Masterpräparationen mit unterschiedlichen okklusalen Reduktionen (0,5 mm, 1 mm, 1,5 mm) durchgeführt. Die Präparation erfolgte non-retentiv, ausschließlich okklusal und mit zwei diagonalen Noppen zur sicheren Positionierung der Restaurationen. Von den drei Masterstümpfen wurden im Anschluss Silikonabformungen hergestellt, in die ein hochgefülltes, lichthärtendes Kompositmaterial schichtweise eingebracht wurde. Alle Restaurationen wurden digital konstruiert (CAD) und maschinell gefertigt (CAM). Die Designs der Zirkoniumoxid-Veneers wurden in der Übergangszone zwischen der basalen 4Y-PSZ-Schicht und der okklusalen 5Y-PSZ-Schicht des mehrschichtigen Rohlings positioniert. Die Positionierung erfolgte standardisiert mit einem Abstand von 3,3 mm von der oberen Rohlingkante zur okklusalen Oberfläche der Restauration. Die Vorbehandlung der Veneers und der dentinanalogen Kompositstümpfe erfolgte gemäß der jeweiligen Herstellerangaben. Im Anschluss wurden die Veneers volladhäsiv mit einem dualhärtenden Kompositzement auf den Kompositstümpfen befestigt. Alle Proben

wurden einer thermodynamischen Belastung unterzogen (49 N, 1,6 Hz, 1,2 Millionen Zyklen, Temperaturwechsel 5-55 °C) und es erfolgte ein Einzelzahn-Belastungstest bis zur Fraktur. Die Frakturanalyse wurde lichtmikroskopisch durchgeführt und klassifikatorisch in vier Versagensmodi eingeteilt. Die statistische Auswertung erfolgte mittels ANOVA, Tukey-Post-hoc-Test und t-Test (Signifikanzniveau $p < 0,05$).

5.4.3 Ergebnisse

Nach der thermodynamischen Belastung der okklusalen Veneers war die Gesamtüberlebensrate 100 %. Es trat kein Debonding und keine Fraktur auf. Jedoch zeigten sechs von 12 Veneers der Zirkoniumoxidgruppe in 0,5 mm Rissbildungen. Damit lag die Gesamt-Erfolgsrate bei 91,7 %. Im Bruchlasttest schnitten LDS-Veneers in allen Schichtstärken signifikant besser ab als die entsprechenden kompositionsgradienten, mehrschichtigen Zirkoniumoxid-Veneers ($p < 0,001$). Die mittleren Bruchlasten zeigten folgendes Ranking: LDS-0,5 mm: 3194 N, LDS-1,0 mm: 2683 N, LDS-1,5 mm: 2338 N, Z-1,5 mm: 1744 N, Z-0,5 mm: 1310 N, Z-1,0 mm: 1198 N. Dabei bestand für alle Schichtstärken ein signifikanter Unterschied zugunsten von LDS. In der LDS-Gruppe war die ultradünne Variante (0,5 mm) signifikant belastbarer als die 1,5-mm-Variante ($p = 0,046$). Bei Zirkoniumoxid waren im Gegensatz zu LDS dickere Veneers (1,5 mm) signifikant stabiler als die dünneren Varianten (1,0 mm: $p = 0,003$, 0,5 mm: $p = 0,02$). Als Hauptversagensmodi wurden bei den LDS-Proben in 0,5 mm Schichtstärke hauptsächlich katastrophale Frakturen mit Wurzelbeteiligung festgestellt, während bei Zirkoniumoxidproben gleicher Schichtstärke vorrangig Frakturen der Restauration und des darunterliegenden Stumpfes auftraten.

5.4.4 Diskussion

Die Ergebnisse der vorliegenden Studie zeigen, dass sowohl das Restaurationsmaterial als auch die Schichtstärke einen signifikanten Einfluss auf die Bruchlast minimalinvasiver okklusaler Veneers haben. Veneers aus LDS wiesen in allen untersuchten Schichtstärken signifikant höhere Bruchlasten verglichen mit solchen aus mehrschichtigem Zirkoniumoxid auf. Besonders hervorzuheben ist die überlegene Belastbarkeit der ultradünnen LDS-Veneers mit 0,5 mm Schichtstärke, die im Mittel die höchsten Bruchlastwerte erzielten. Die Positionierung der Zirkoniumoxid-Veneers in der Übergangszone zwischen 4Y- und 5Y-PSZ stellt einen bislang kaum

untersuchten Aspekt dar. Die vorliegende Studie zeigte, dass hinsichtlich der Bruchlast die Positionierung in dieser Zone keine mechanisch verstärkende Wirkung im Vergleich zu LDS darstellt.

Die hohen Bruchlastwerte der 0,5 mm dünnen LDS Veneers lassen sich durch das Zusammenspiel von Materialeigenschaften, Schichtstärke und der Biomechanik des Zahn-Restoration-Komplexes erklären. Neben der höheren Gesamtstabilität durch den Erhalt größerer Anteile natürlicher Zahnschubstanz können sich in der Schichtstärke reduzierte Veneers unter Kaubelastung stärker elastisch verformen, wodurch an der Kontaktfläche zum Zahn und entlang der Klebefuge vermehrt Druckspannungen entstehen. Diese hemmen die Rissausbreitung und tragen so zur höheren Bruchlast der Restauration bei [141]. Zudem weist LDS im Vergleich zu Zirkoniumoxid aufgrund seines zahnschubstanzähnlicheren Elastizitätsmoduls (Schmelz ~70 GPa, Dentin ~18, LDS: ~95 GPa, Zirkoniumoxid: ~210 GPa) geringere Spannungsunterschiede im Bereich des Übergangs der Restauration zur Zementschicht auf, wodurch das Risiko für Biegefrakturen reduziert wird [98].

Abstrahlen ist für die adhäsive Vorbehandlung von Zirkoniumoxid unerlässlich, da herkömmliche silanbasierte Verfahren keine ausreichende Haftung erzielen [104, 142]. Zu hohe Strahlparameter können jedoch Mikrorisse erzeugen, die sich unter Kaubelastung fortsetzen und die Stabilität beeinträchtigen können [143, 144]. Auch bei reduzierten Restaurationen aus 5Y-PSZ mit geringerem Materialwiderstand könnte das vorherige Abstrahlen eine Rissinitiierung und Ausbreitung begünstigen haben. Eine andere Studie berichtet ebenfalls über Rissbildungen in okklusalen Veneer-Restaurationen aus 5Y-PSZ [145], was die Empfindlichkeit dieses Materials unterstreicht. Ein Abstrahlprotokoll von 50–60 µm Partikelgröße und max. 2 bar Druck sollte daher strikt eingehalten werden [146]. Zudem ist eine verlässliche adhäsive Befestigung nur dann gewährleistet, wenn Oberflächenvorbehandlung und Befestigungskomposit optimal auf das Material abgestimmt sind [142]. Um die Ergebnisse dieser Studie umfassender interpretieren zu können, ist eine detaillierte Frakturanalyse der Rissverläufe und Initiationspunkte erforderlich.

5.5 Originalarbeit 5

Failure prediction of lithium disilicate and composition-gradient multilayered zirconia occlusal veneers: A fractographic and theoretical analysis

5.5.1 Hintergrund

Basierend auf den Ergebnissen der unter 5.4 beschriebenen experimentellen in vitro Studie war das Ziel der vorliegenden Arbeit, die unter Ermüdungsbelastung beobachteten Rissbildungen fraktographisch zu analysieren. Ergänzend wurde zur Abschätzung der Lebensdauer der Restaurationen ein mechanisches Modell herangezogen, das an die Biegebeanspruchung dünner Platten auf elastischer Unterlage angelehnt ist. Diese ergänzende Untersuchung wurde durchgeführt, da die experimentellen Daten allein keine Rückschlüsse auf die mechanischen Ursachen der Rissinitiierung sowie die Lebensdauer dieser Restaurationen zulassen.

5.5.2 Material und Methode

In der vorliegenden Arbeit wurde eine fraktographische Analyse jener Proben durchgeführt, bei denen nach thermomechanischer Ermüdungsbelastung (Abschnitt 5.4.3) Rissbildungen auftraten. Hierzu wurden die betroffenen Proben zunächst lichtmikroskopisch untersucht und anschließend mittels REM analysiert. Ergänzende Aufnahmen im Rückstreuелеktronenmodus (BSE) ermöglichten eine differenzierte Darstellung materialabhängiger Kontraste und Rissverläufe an der Restaurationsoberfläche. Zur theoretischen Abschätzung der Lebensdauer wurde ein rechnerisches Modell herangezogen, das von einer Biegebeanspruchung okklusaler Veneer-Restaurationen auf einem nachgiebigen Substrat ausgeht. Auf Grundlage werkstoffspezifischer Parameter (Biegefestigkeit, Elastizitätsmodul, Oberflächenrisslänge und Rissausbreitungsverhalten) wurde der kritische Belastungszustand bestimmt und die Lebensdauer bis zum strukturellen Versagen in Lastzyklen berechnet. Diese Modellierung erlaubte eine quantitative Bewertung der Versagenswahrscheinlichkeit in Abhängigkeit von Material und Schichtstärke und diente dem Vergleich mit den experimentellen Ergebnissen.

5.5.3 Ergebnisse

Die fraktographische Analyse identifizierte die Rissinitiierung an der Unterseite der Veneers in Folge von Zugspannungen als Versagensursache. Die Modellberechnung zur Simulation der Biegebeanspruchung auf elastischem Substrat ergab für 0,5 mm 5Y-PSZ-Veneers eine hohe Versagenswahrscheinlichkeit nach $<10^6$ Zyklen. Bei Schichtstärken $<0,7$ mm stieg das Risiko radialer Frakturen deutlich an. Für LDS-Veneers in 0,5 mm Schichtstärke wurde eine rechnerische Lebensdauer von bis zu 10^{10} Zyklen ermittelt. Ab 0,7 mm Schichtstärke erreichten auch Zirkoniumoxid-Veneers Lebensdauern im Bereich von LDS.

5.5.4 Diskussion

Die Ergebnisse der theoretischen Berechnungen bestätigen, dass ultradünne (0,5 mm) okklusale Veneers aus LDS eine deutlich höhere Ermüdungsbeständigkeit aufweisen als solche aus mehrschichtigem Zirkoniumoxid. Ab einer Schichtstärke von 0,7 mm erreichten Zirkoniumoxid-Veneers rechnerisch eine Lebensdauer, die mit jener von LDS vergleichbar oder sogar überlegen war. Bei Zirkoniumoxid scheint eine erhöhte Schichtstärke die durch okklusale Belastungen erzeugten Zugspannungen an der Unterseite der Restauration zu reduzieren, welche maßgeblich für das Auftreten radialer Biegefrakturen verantwortlich sind. Gleichzeitig verringert sich die Durchbiegung des Materials, wodurch weniger kritische Spannungen in die Zementierungszone übertragen werden.

Ein zentraler Unterschied zwischen beiden Materialien ergibt sich aus ihren unterschiedlichen Elastizitätsmodulen. Das geringere Elastizitätsmodul von LDS (~95 GPa) führt im Vergleich zu Zirkoniumoxid (~210 GPa) zu einer besseren Anpassung an die elastischen Eigenschaften des dentinanalogen Untergrundes (~18 GPa) [98]. Diese höhere Nachgiebigkeit ermöglicht eine gleichmäßigere Spannungsverteilung und reduziert lokale Spannungsspitzen, die als Ausgangspunkte für Rissinitiierungen wirken können [132, 147]. Bei Zirkoniumoxid hingegen bewirkt die höhere Steifigkeit eine verstärkte Spannungsakkumulation an der Unterseite der Restauration, was die beobachtete Rissanfälligkeit bei geringer Schichtstärke erklären könnte. Klinisch relevant ist zudem, dass ultradünne LDS-Veneers meist auf natürlichem Schmelz adhäsiv befestigt werden, dessen Elastizitätsmodul (~70 GPa) dem von LDS (~95

GPa) deutlich näher liegt als dem von Zirkoniumoxid [148]. Diese elastische Kompatibilität begünstigt eine homogene Kraftübertragung und kann zu einer noch besseren Ermüdungsbeständigkeit dünner LDS-Veneers in klinischen Settings führen [149].

Die im experimentellen Teil beobachteten radialen Rissbildungen bei 0,5 mm dicken Zirkoniumoxid-Veneers konnten durch die fraktographische Analyse eindeutig auf die Rissinitiierung an der Zementierungsfläche zurückgeführt werden. Aus klinischer Sicht gelten radiale Frakturen als besonders problematisch, da sie unterhalb der okklusalen Oberfläche entstehen und meist nicht sichtbar sind [150]. Dies birgt das Risiko eines verzögerten, katastrophalen Versagens unter funktioneller Belastung [98]. Eine ausreichende Schichtstärke von $<0,7$ mm erscheint für Zirkoniumoxid daher essentiell, um das Risiko radialer Frakturen zu minimieren. In Situationen, in denen eine hohe mechanische Belastbarkeit im Vordergrund steht und ästhetische Anforderungen eine untergeordnete Rolle spielen, wie etwa im Molarenbereich, kann monolithisches Zirkoniumoxid (3Y-TZP oder 4Y-PSZ) oder LDS gegenüber transluzenten mehrschichtigen Zirkoniumoxid-Materialien bevorzugt werden. In Regionen mit hohen ästhetischen Anforderungen bieten okklusale Veneers aus mehrschichtigen Zirkoniumoxid hingegen eine vielversprechende restaurative Option, solange eine ausreichende Materialstärke von $>0,7$ mm eingehalten wird.

6. Diskussion

Studien, die digitale und konventionelle Workflows miteinander vergleichen, zeigen heterogene Ergebnisse und verdeutlichen den Bedarf weiterer Optimierungen des digitalen Workflows. Systematische Übersichtsarbeiten berichteten einerseits von vergleichbarer Genauigkeit, Passung, Bruchfestigkeit sowie Langzeit-Performance beider Workflows [151, 152]. Im Gegensatz dazu zeigte eine andere systematische Übersichtsarbeit ein 1,84-fach erhöhtes Versagensrisiko von CAD/CAM gefertigten festsitzenden Restaurationen (Einzelrestaurationen sowie Brücken) im Vergleich zu konventionellen Arbeiten [153]. Dabei ist jedoch zu berücksichtigen, dass die in die Analyse eingeschlossenen Studien einen Zeitraum von 1999 bis 2017 abdecken und somit auch frühe CAD/CAM-Systeme, Materialien und Fertigungsprozesse einschließen, die nicht mit den heutigen digitalen Workflows vergleichbar sind. Eine aktuelle in vitro Studie zu implantatgetragenen, vollständig digital gefertigten Titan-Gerüsten weist ebenfalls auf Limitationen des digitalen Workflows hin, da bei bis zu 80 % der Gerüste signifikante Passungenauigkeiten festgestellt wurden [35]. Eine in vitro Untersuchung zur Fehlerakkumulation innerhalb des digitalen Workflows (Datenerfassung mit IOS, CAD, CAM) zeigte, dass die digitale Abformung die größte Fehlerquelle darstellt, während die CAD-Phase im Vergleich zur digitalen Abformung und Fertigung die geringsten Abweichungen aufwies [18]. Die in dieser Habilitationsschrift eingeschlossenen Untersuchungen befassen sich daher gezielt mit den Einflussfaktoren dieser beiden Teilbereiche des digitalen Workflows und deren Optimierungsmöglichkeiten.

In der Literatur werden verschiedene Methoden zur Bewertung der Genauigkeit verwendet, was die Ergebnisse beeinflussen und die Vergleichbarkeiten zwischen einzelnen Studien erschweren kann. Häufig wird die Genauigkeit der digitalen Abformung mittels Überlagerung der Datensätze in einem 3D-Evaluationsprogramm analysiert [36, 54]. Nachteilig ist hierbei, dass auf diese Weise lediglich ein einzelner Schritt und nicht der gesamte digitale Workflow bewertet wird. Studien, die den gesamten Workflow zur Herstellung implantatgetragener Restaurationen evaluieren, verwenden häufig den Ein-Schrauben-Test (Sheffield-Test), Schraubenwiderstandstest, radiographische Verfahren oder simulieren digitale Schraubentests

innerhalb einer CAD-Software [35, 154]. Derzeit besteht keine Einigkeit darüber, welche Abweichungen als klinisch akzeptabel gelten. Für die Evaluation implantatbasierter Ganzkieferabformungen werden lineare Abweichungen zwischen 50-150 µm sowie angulare Abweichungen zwischen 0,4-1° als akzeptabel beschrieben [39]. Eine klinisch akzeptable Passung implantatgetragener Gerüste wird häufig mit einer Toleranzgrenze von 150 µm definiert [155, 156]. Für festsitzende Restaurationen wird die klinisch akzeptable Randpassung je nach Quelle mit Werten zwischen 120 und 200 µm angegeben [157, 158]. Limitierend ist, dass viele Studien ihre festgestellten Abweichungen häufig gar nicht bezüglich einer Schwelle zur klinischen Relevanz einordnen.

Für digitale Ganzkieferabformungen werden in klinischen Studien Abweichungen zwischen 73 und 433 µm (Richtigkeit) und 80 und 199 µm (Präzision) beschrieben [27, 159]. Diese können vor allem bei verblockten implantatgetragenen Ganzkieferrekonstruktionen zu Fehlpassungen mit Schraubenlockerungen bis hin zu Frakturen der Schraube oder des Implantates führen [54]. Die erste eingeschlossene Studie konnte zeigen, dass PG die Passung implantatgetragener Ganzkiefer-Restaurationen verbessern kann. Die Überlegenheit von PG-Systemen deckt sich mit aktuellen Übersichtsarbeiten, die ebenfalls eine höhere Genauigkeit der digitalen Abformung von PG im Vergleich zu IOS für implantatgetragene Ganzkieferversorgung berichten [53, 160]. Die vorliegende Studie konnte jedoch erstmals zeigen, dass PG-Systeme nicht nur auf Abformebene, sondern auch über den gesamten Workflow eine höhere Genauigkeit erzielen als IOS. Beim intraoralen Scan werden die Scanbodies als Videosequenz erfasst und erscheinen in der finalen Abformung. Bei der PG-Abformung hingegen werden die Marker bereits automatisch durch standardisierte Library-Elemente ersetzt. Inwieweit sich diese Unterschiede auf die Genauigkeit im Gesamtworkflow auswirken, ist bisher wenig bekannt. Zudem sind PG-Systeme nicht in der Lage Weichgewebe zu erfassen, sodass die Abformung mit einem intraoralen Scan gematcht werden muss [50], was eine weitere Fehlerquelle darstellen kann. Die in dieser Arbeit eingeschlossene erste Studie bestätigt, dass nicht nur die eigentliche PG-Datenerfassung, sondern auch die in der CAD-Phase auftretenden Abweichungen innerhalb eines klinisch akzeptablen Bereichs lagen. Zudem konnten in der anschließenden CAM-Fertigung keine zusätzlichen Fehlerquellen relevanter Größenordnung festgestellt werden, die zu einer

nennenswerten additiven Fehlerakkumulation über die Prozessschritte hinweggeführt hätten.

Die Ergebnisse drei weiterer eingeschlossener Arbeiten verdeutlichen, dass innerhalb des digitalen Workflows nicht nur die digitale Abformung, sondern eine geeignete Kombination aus Keramikmaterial, Restaurationsdesign und Schichtstärke entscheidend für die mechanische Leistungsfähigkeit vollkeramischer Versorgungen ist. In der vierten Studie konnte für okklusale Veneers der Einfluss des Keramikmaterials (LDS > mehrschichtiges Zirkoniumoxid [4Y,5Y-PSZ]) in allen getesteten Schichtstärken nachgewiesen werden. Für minimalinvasive Kronen in 0,5 mm Schichtstärke lagen die Bruchlastwerte von 3Y-TZP Zirkoniumoxid (1122 N) und LDS (1193 N) in zwei weiteren eingeschlossenen Studien auf einem ähnlichen Niveau. Die Vergleichbarkeit der beiden Untersuchungen ist aufgrund der unterschiedlichen Belastungsprotokolle (200 N vs. 49 N) zwar eingeschränkt, jedoch bestätigen Literaturdaten die gewonnene Beobachtung. Sie berichten, dass LDS-Restaurationen die mechanische Leistungsfähigkeit von 5Y-PSZ Zirkoniumoxid übertreffen und bei geeigneter adhäsiver Befestigung sogar vergleichbare Werte zu 3Y-TZP erreichen können [98, 161].

In allen drei Studien wurde ein signifikanter Einfluss der Schichtstärke festgestellt, wobei die Richtung des Effektes vom Restaurationsdesign und Keramikmaterial abzuhängen scheint. Während bei Zirkoniumoxidrestaurationen die Erhöhung der Schichtstärke sowohl für Kronen als auch okklusale Veneers mit einer Erhöhung der Bruchlast einherging, zeigten okklusale Veneers aus LDS genau den gegenteiligen Effekt (reduzierte LDS Veneers waren stabiler). In der Literatur wird die Kombination aus dem geringeren Elastizitätsmodulunterschied zwischen LDS und der Zahnhartsubstanz sowie einer besseren Widerstandsfähigkeit des Zahn-Restaurations-Komplexes als Ursache für die überlegene mechanische Stabilität von minimalinvasiven LDS-Restaurationen vermutet [98, 149]. Diese Ergebnisse decken sich mit anderen Studien, die ebenfalls einen vom Material und Restaurationsdesign abhängigen Einfluss der Schichtstärke auf das Bruchlastverhalten von teilbedeckenden Restaurationen feststellten [132, 162]. Für Kronen wurde hingegen in mehreren Studien kein signifikanter Einfluss der Schichtstärke festgestellt [163, 164]. Die Kombination aus reduzierter Schichtstärke, minimalinvasivem

Präparationsdesign und die Wahl eines geeigneten Keramikmaterials, wie etwa LDS, scheint sich besonders vorteilhaft auf die mechanische Stabilität der Restauration auszuwirken. Bislang fehlten jedoch in vitro Studien, die das Ermüdungsverhalten von verschiedenen minimalinvasiven Restaurationsformen vergleichen. Die dritte eingeschlossene Studie konnte für den direkten Vergleich von Kronen und Teilkronen aus LDS eine signifikante Überlegenheit des minimalinvasiveren Teilkronendesigns feststellen. Zudem zeigen die Bruchlastwerte der dritten und vierten Studie für LDS in 0,5 mm Schichtstärke eine deutliche Überlegenheit minimalinvasiver Restaurationsformen (Kronen: 1193 N, Teilkronen: 2023 N, okklusale Veneers: 3194 N). Ursächlich scheint neben dem Substanzerhalt auch eine günstigere Spannungsverteilung zu sein. Durch die zirkuläre Umfassung der Zahnschubstanz bei der Vollkrone entsteht ein vergleichsweise starres Restaurations-Zahn-System. Dies kann zu einer weniger gleichmäßigen Spannungsverteilung unter funktioneller Belastung führen und dadurch die Ausbildung lokaler Spannungsspitzen begünstigen [165].

In klinischen Studien werden unterschiedliche minimalinvasive Schichtstärken nur selten systematisch untersucht [74]. Untersuchungen, die unterschiedliche Restaurationsdesigns vergleichen, liefern keine Information zu deren Schichtstärke [139]. Eine systematische Übersichtsarbeit basierend auf sechs klinischen Kurzzeitstudien (Beobachtungszeit 1-3 Jahre) stellte einen gleichwertig hohen Erfolg für Kronen und Teilkronen, hergestellt aus verschiedenen Keramikmaterialien, fest [166]. Auch in klinischen Langzeitstudien wiesen sowohl Kronen als auch Teilkronen/Onlays aus LDS hohe Überlebensraten (80-100 % nach 13-15 Jahren) auf [101, 135, 139]. Aufgrund der Zahnschubstanzschonung wurde jedoch empfohlen, Teilkronen gegenüber Vollkronen für kleinere Defekte zu bevorzugen [139]. Der Erhalt größerer Anteile der Zahnschubstanz bietet günstigere Voraussetzungen für eine spätere Neuversorgung. Darüber hinaus gewährleistet der Erhalt des natürlichen Zahnschmelzes einen besseren adhäsiven Verbund, da die Phosphorsäureätzung ein an die Schmelzprismen angepasstes Mikroporositäten-Relief erzeugt, in das Monomere tief eindringen und stabile „Tags“ ausbilden können [167]. Dentin weist dagegen eine feuchte tubuläre, kollagenbasierte Matrix mit Smear Layer und einer hydrolyse- sowie matrixmetalloproteinaseanfälligen Hybridzone auf, was die Langzeit-Haftfestigkeit limitiert [168, 169]. Die Identifikation von Initiationspunkten und

Rissverläufen in der fünften Studie verdeutlichte ebenfalls, dass bei dünnwandigen Restaurationen nicht allein materialbedingte Grenzen, sondern vor allem die Interaktionen von Keramikmaterial, Restaurationsdesign, Schichtstärke, adhäsiver Befestigung und funktioneller Belastung entscheidend sind. Die Studie konnte eine Mindestschichtstärke von 0,7 mm für okklusale Veneers aus mehrschichtigem Zirkoniumoxid (4Y- und 5Y-PSZ) ableiten. Übereinstimmend damit zeigte ein Review, dass okklusale Veneers bei Schichtstärken von $\geq 0,7$ mm bei adäquatem Design und korrekt angewandter Adhäsivtechnik eine ausreichende mechanische Performance erreichen können [71]. Mehrschichtige Zirkonoxidkeramiken wurden jedoch nicht explizit als eigenständige Materialklasse berücksichtigt. Ihre Prognose bleibt bei reduzierten Schichtstärken weiterhin unzureichend untersucht [145] und weitere präklinische und klinische Studien sind notwendig.

7. Zusammenfassung und Ausblick

Die folgenden wissenschaftlichen Fragestellungen sollten mit den eingeschlossenen Originalarbeiten beantwortet werden:

- 1.) Wie wirkt sich die Wahl unterschiedlicher digitaler Abformverfahren (PG vs. IOS) auf die Genauigkeit implantatgetragener Ganzkieferrekonstruktionen aus?
- 2.) Inwieweit beeinflussen Materialauswahl, Restaurationsdesign und reduzierte Schichtstärke die mechanische Leistungsfähigkeit keramischer Versorgungen innerhalb des digitalen Workflows?

Die Ergebnisse zeigen, dass digitale Workflows in der restaurativen Zahnmedizin bereits zuverlässige Resultate ermöglichen. Für implantatgetragene Ganzkieferrestaurationen erwies sich die PG als besonders präzise und zeigte geringere Abweichungen über den gesamten Workflow. IOS konnten diese Indikation zwar ebenfalls mit klinisch akzeptabler Genauigkeit abdecken, zeigten aber eine höhere Fehleranfälligkeit im Vergleich zur PG. Damit lässt sich feststellen, dass beide Systeme ihren Stellenwert im digitalen Workflow besitzen: Ein spezifischer Vorteil der PG ergibt sich insbesondere für Praxen, die regelmäßig großspannige, implantatgetragene Versorgungen bei vollständig zahnlosen Patienten durchführen. In diesen Situationen kann die höhere Genauigkeit den Anpassungsaufwand

reduzieren und die Effizienz des Workflows verbessern. Für Praxen, in denen lediglich kleinere oder weniger komplexe implantatgetragene Restaurationen angefertigt werden, stellt der IOS eine ökonomisch sinnvolle Lösung dar.

Für keramische Einzelzahnrestaurationen zeigte sich, dass die mechanische Leistungsfähigkeit der Restauration nicht primär von der Festigkeit des Keramikmaterials abhängt, sondern maßgeblich durch die Kombination aus Keramikmaterial, Restaurationsdesign und Schichtstärke beeinflusst wird. Minimalinvasive Designs wie Teilkronen und okklusale Veneers wiesen bei gleicher Schichtstärke (0,5 mm) eine höhere Bruchfestigkeit auf als Vollkronen. Die Verbindung aus reduzierter Schichtstärke, minimalinvasivem Restaurationsdesign und dem Einsatz eines geeigneten Keramikmaterials, wie LDS, erwies sich als besonders günstig. Für okklusale Veneers aus mehrschichtigem Zirkoniumoxid konnte zudem eine Mindestschichtstärke von 0,7 mm abgeleitet werden, was zeigt, dass sich durch gezielte Abstimmung von Material, Design und Schichtstärke klinisch zuverlässige Versorgung im digitalen Workflow realisieren lassen.

Die Leistungsfähigkeit mehrschichtiger Zirkoniumoxidkeramiken in reduzierten Schichtstärken sollte in weiteren in vitro Studien untersucht sowie deren klinische Übertragbarkeit in zukünftigen klinischen Studien validiert werden. Hybride Akquisitions-Strategien, welche PG und IOS kombinieren, können Fortschritte in Genauigkeit und Effizienz ermöglichen. Auch der Einsatz von KI eröffnet neue Perspektiven: Sie könnte zukünftig die Analyse der Abformdaten und das Designen der Restaurationen automatisieren [170]. Roboter-Technologien repräsentieren den nächsten Schritt, indem sie den gesamten digitalen Workflow automatisieren und sowohl Behandlungszeit als auch menschliche Fehlerquellen weiter minimieren [171]. Im Roboter-Workflow wird die Präparation digital vorgeplant und anschließend durch den Präparationsroboter ausgeführt. Basierend auf Datensätzen der Präparationssimulation ist es bereits möglich, Restaurationen vor der eigentlichen Präparation am Patienten herzustellen und innerhalb derselben Behandlungssitzung zu zementieren [172]. Erste präklinische und klinische Studien belegen die Durchführbarkeit robotergestützter Präparationssysteme: Semiaktive robotische Systeme erzielten bei der Präparation von Keramikveneers klinisch akzeptable Genauigkeiten, wenngleich diese jedoch unterhalb der Genauigkeit erfahrener

Behandler lagen [173]. In der ersten klinischen Machbarkeitsstudie mit einem teilautomatisierten, robotergestützten Präparationssystem konnten Einzelkronen mit einer Genauigkeit von unter 50 µm in nur einer Sitzung hergestellt werden [172]. Vor diesem Hintergrund stellen robotergestützte Präparationssysteme bereits heute einen vielversprechenden Ansatz dar, um minimalinvasive Präparationen standardisiert, reproduzierbar und weitgehend unabhängig von der individuellen Erfahrung des Behandlers umzusetzen. Dennoch weisen KI- und robotergestützte Systeme derzeit noch relevante Limitationen auf, insbesondere hinsichtlich der begrenzten klinischen Datenlage, der Abhängigkeit von standardisierten Datensätzen sowie der bislang eingeschränkten Adaptationsfähigkeit an individuelle klinische Situationen [174]. Auch technische Restriktionen, wie eine begrenzte Sensitivität optischer Erfassungssysteme sowie algorithmische Ungenauigkeiten, können zu Abweichungen in Datensätzen führen [173]. Potentielle Fehlkalibrierungen und eingeschränkte Interpretierbarkeit von Entscheidungsprozessen stellen weitere Herausforderungen der KI-Anwendung dar [175]. Während die Ergebnisse diagnostischer Modelle unmittelbar mit anderen diagnostischen Maßnahmen verglichen werden können, lässt sich eine KI-erzeugte Prognose nicht sofort validieren. KI sollte daher als unterstützendes Instrument verstanden werden, das die fachliche Expertise des Zahnarztes ergänzt und Arbeitsabläufe effizienter gestaltet, während die abschließende Beurteilung und Entscheidungsfindung weiterhin in der Verantwortung des Behandlers liegen.

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10. Eidesstattliche Erklärung

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Düsseldorf, 10.02.2026

11. Anhang Originalarbeiten

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Für die Aufnahme der Originalarbeiten aus dem *International Journal of Prosthodontics* sowie dem *Journal of Prosthodontic Research* liegen entsprechende schriftliche Genehmigungen vor.

In Vitro Comparison Between Complete-Arch Implant Supported Frameworks Using Intraoral Scanning and Photogrammetry

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Abstract

Purpose. To compare the vertical microgap of complete arch implant supported frameworks (ISF) obtained from one intraoral scanner (IOS) and two different photogrammetry (PG) systems. **Materials and methods.** An edentulous mandibular model with four implants (BioHorizons) served as reference. Multi-unit abutments (BioHorizons) were inserted and

torqued to 30 Ncm. A digital reference cast (Control group) was obtained by scanning the model with a laboratory scanner (inEosX5, Dentsply Sirona). Three test groups were investigated: PS (Primescan, Dentsply Sirona), ICam (ICam4D, Imetric4D Imaging Sàrl), and PIC (PIC System, PIC Dental). All scans were exported as Standard-tessellation language (STL) files and virtual bars were designed and milled out of titanium (n=2 per test group). The modified Sheffield test was used to quantitatively assess vertical microgaps. Differences between all groups were analyzed using Kruskal-Wallis test ($\alpha=0.05$). Analysis of variance (ANOVA) was used to compare vertical microgaps at each respective implant position across groups ($\alpha=0.05$). Results. Mean vertical microgaps varied across acquisition devices, with lowest values in the control group ($22.65 \pm 7.279 \mu\text{m}$), followed by ICam ($24.10 \pm 7.991 \mu\text{m}$), PIC ($30.52 \pm 9.809 \mu\text{m}$), and PS ($33.21 \pm 13.84 \mu\text{m}$). Statistically significant differences were detected between Control and PS as well as Control and ICam. No statistically significant difference occurred between the two PG groups. Vertical microgaps were the highest at implant site 44, predominantly observed in groups PS and PIC. Conclusions. Frameworks fabricated from digital impressions of PG groups showed lower vertical microgaps than those fabricated using the IOS. *Int J Prosthodont* 2025. doi: 10.11607/ijp.9363

Introduction

Computer-aided design and computer-aided manufacturing (CAD-CAM) technology has gained popularity in restorative dentistry, because it reduces human error, improves time and cost efficiency as well as the patient comfort.¹⁻³ Whether the digital workflow is superior for the fabrication of complete arch implant supported frameworks (ISF), however, still remains a matter of debate.⁴ The amount of errors when scanning edentulous arches with multiple implants depends on several factors. These include the inadequate presence of anatomical

landmarks, algorithm employed for image stitching, type and material of scan bodies, scanning distance, scanning strategy as well as environmental and patient-related factors.⁵⁻⁷ Especially, when a larger number of images are captured, algorithms used to stitch them together can introduce errors during the 3D reconstruction process.⁸ Previous studies have attempted to improve the accuracy of digital implant impressions by investigating various splinting methods (e.g., dental floss, acrylic resin, calibrated metal frameworks,^{9,10} anatomic landmarks,¹¹ scanning strategies,^{11,12} or alternative acquisition methods, such as photogrammetry (PG),^{13,14} however, results have been controversial.

PG is a non-contact imaging technique that uses photographs to create 3D models.¹⁵ It was introduced to restorative dentistry 30 years ago to detect framework misfit in dental implants.^{16,17} Recent advances in digital photography and the development of specialized software have transformed PG into an accurate, reliable, and cost-effective method for the fabrication of implant-supported prostheses.¹⁵ Unlike intraoral scanners (IOS), which require an image collection of the surfaces of scan bodies and soft tissues, PG systems are not influenced by ambient light or less structured edentulous areas.^{18,19} Although PG requires capturing only a few target points on custom scan bodies, it is unable to simultaneously capture soft tissues, hard tissues, and implant positions.^{9,14} As a result, PG is often combined with an IOS to enhance the accuracy of digital implant impressions in edentulous arches. Although previous studies have confirmed that PG systems can accurately transfer intraoral implant positions to a virtual model, findings have been inconsistent across different studies.^{9,20,21} Moreover, the application and influence of PG on the entire workflow is limited to case reports.^{22,23} Therefore, the present in vitro study aims to evaluate and compare vertical microgaps of milled ISF fabricated from digital impressions of one IOS and two different PG systems. The null hypothesis was that there was no statistically significant difference in vertical microgaps of ISF obtained from different acquisition techniques.

Materials and Methods

Four implants (Tapered Internal Plus, 3.5 mm Platform Implant, BioHorizons) were placed non-guided and parallel to each other into an edentulous model with type 2 bone density, adhering to the manufacturer's surgical protocol (Fig 1A). Implant positions (32, 42 [mandibular lateral incisors] and 34, 44 [mandibular first premolars] according to the World Dental Federation [FDI] System) were chosen to simulate the placement of an implant-supported prosthesis replacing the missing teeth up to the first molars. Multi-unit abutments (MUA, Internal 3.5mm multi-unit abutment, straight, non-hexed, BioHorizons) were inserted into the implants and torqued to 30 Ncm (Fig 1B). Scan bodies (Elos Accurate multi-unit scan body, Elos Medtech) were positioned and hand tightened on each MUA (Fig 2A). A laboratory scanner (inEos X5, Dentsply Sirona) with an accuracy of $2.1 \pm 2.8 \mu\text{m}^{24}$ was used to obtain a digital reference cast, which was exported as Standard Tessellation Language (STL) file and served as control (group Control). Three different test groups were created based on the digital impression technique used: PS (Primescan, Dentsply Sirona), ICam, (ICam4D, Imetric4D Imaging Sàrl) and PIC (PIC System, PIC Dental) (Fig 3). In group PS, scan bodies (Elos Accurate multi-unit scan body, Elos Medtech) were placed and hand tightened on each MUA (Fig 2A) and two abutment-level scans (group PS, n=2) were performed with the IOS (Primescan, Dentsply Sirona). In the PG groups, optical markers (ICamBody, Imetric 4D Imaging Sàrl; PIC transfer, PIC Dental) (Fig 2B and 2C) were hand-tightened onto the MUAs and digitized using PG-cameras (ICamBody, Imetric 4D Imaging Sàrl; PIC System, PIC Dental, n=2 per group). Scans of all test groups were exported as STL-files and sent to an authorized milling center (Vulcan Custom Dental). Titanium (Ti-6Al-4V ELI titanium alloy) bars were milled using a 5-axis milling machine (CORiTEC 650i, imes-icore) and afterwards marked for a randomized examination (Control group: n=1; test groups: n=2) (Fig 4). To mitigate bias, a blinded investigator, unaware of the group allocation,

performed all measurements. The modified Sheffield test was applied for quantitative assessments of vertical microgaps (Fig 5). In the original Sheffield test, a single screw is used to secure the bar onto a distally positioned implant abutment. If the fit is not passive, the bar lifts, resulting in a gap at one or more abutment interfaces.²⁵ In the modified Sheffield test, the bar is initially placed onto all implant abutments with all screws inserted. Starting from the center, each screw is gradually tightened with equal force to seat the bar evenly. Once the bar is fully seated, all screws are removed except for one.²⁶



Fig 1a Edentulous mandibular reference model with four implants (Tapered Internal Plus, 3.5 mm Platform Implant, BioHorizons), (b) Reference model with torqued multi-unit abutments (Internal 3.5mm multi-unit abutment, non-hexed, BioHorizons).

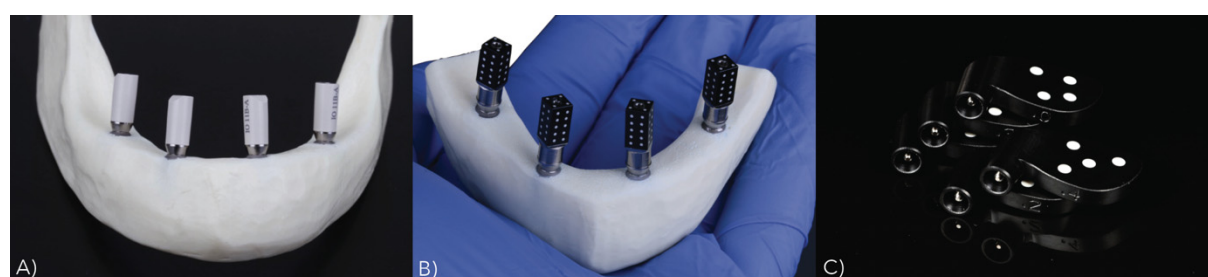


Fig 2 Reference model with (a) Intraoral scan bodies (Elos Accurate multi-unit scan body, Elos Medtech), (b) ICamBodies (IcamBody; Imetric4D Imaging Sàrl), (c) PIC transfers (PIC transfer; PIC Dental).

This peer-reviewed, accepted manuscript will undergo final editing and production prior to print publication.
Any blinded information will be available then.

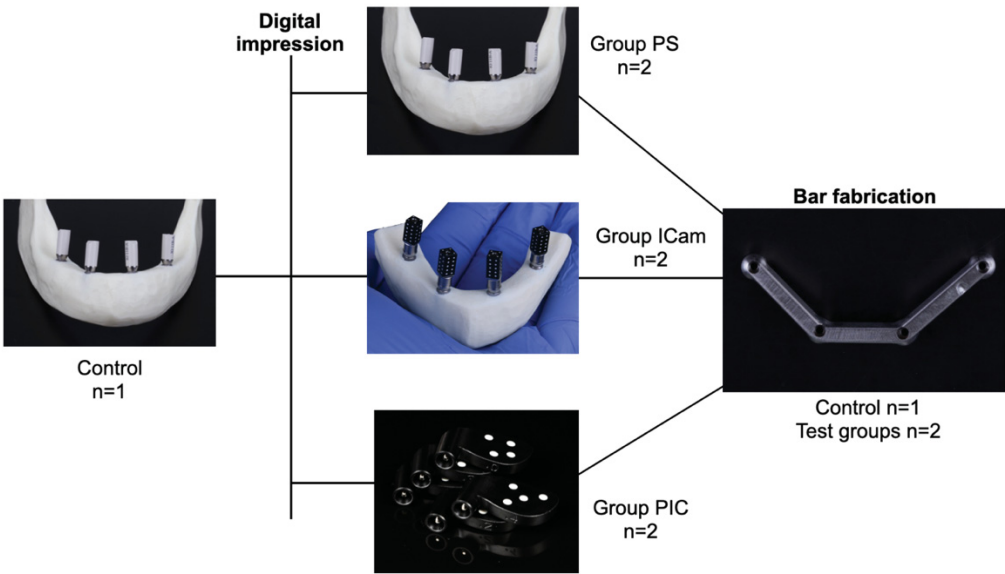


Fig 3 Test setup. Control: inEos X5 (Dentsply Sirona), PS: Primescan (Dentsply Sirona), ICam: Cam4D (Imetric4D), PIC: PIC System (PIC Dental).



Fig 4 Milled titanium bar.



Fig 5 Modified Sheffield Test: All screws were removed except for the one in position 32.

Each Titanium bar was fixed on the MUAs with all screws inserted and torqued to 5 Ncm. Finally, all screws were removed except for one in position 32 and abutment-framework connection gaps were measured for the non-screw retained implants (34, 42, and 44) using a calibrated scanning electron microscope (FEI Quanta 600 FEG Mark II, resolution 1.5 nm). Images of the buccal, mesial, and distal interfaces were taken at a magnification of 1600x and measurements were performed using a digital SEM software caliper (Fig 6). Mean vertical microgaps were calculated from three mesial, three buccal and three distal measurements for each non-engaged implant-bar interface; hence in total for 27 (9 sites x 3 implants) measurements per titanium framework. Statistical analyses were performed using GraphPad Prism (GraphPad Software). Descriptive data on vertical microgaps (mean and SD) were calculated for all groups. Shapiro-Wilk test was used to assess data normality. Differences between all groups were analyzed using Kruskal-Wallis test ($\alpha=0.05$). Analysis of variance (ANOVA) was used to compare vertical microgaps at each respective implant position across groups ($\alpha=0.05$), followed by Tukey post hoc test ($\alpha=0.05$).

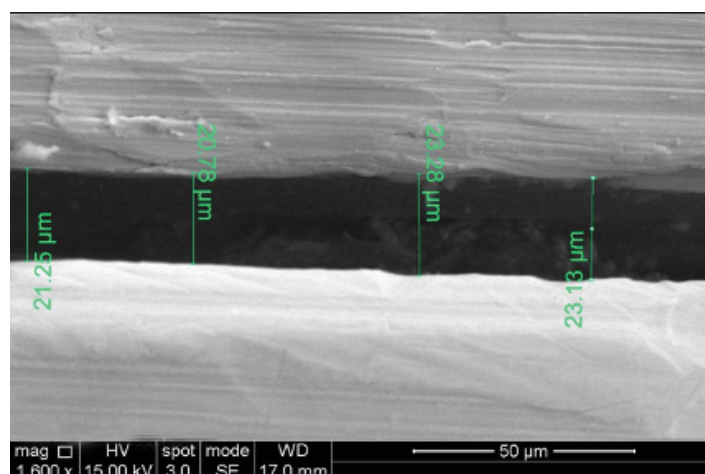


Fig 6 SEM image with digital measurements at the implant-bar interface (magnification 1600x).

Results

Mean vertical microgaps were the lowest in the control group ($22.65 \pm 7.279 \mu\text{m}$), followed by ICam ($24.10 \pm 7.991 \mu\text{m}$), PIC ($30.52 \pm 9.809 \mu\text{m}$), and PS ($33.21 \pm 13.84 \mu\text{m}$). Statistically significant differences were detected between Control and PS ($p=0.0167$) as well as PS and ICam ($p=0.0492$). No statistically significant difference was detected between the two PG groups ($p=0.3439$) (Fig 7). Vertical microgaps were the highest at implant site 44, predominantly observed in groups PS and PIC (Fig 8). The analysis of measurements separated by implant position revealed no statistically significant difference between groups at implant site 34. At implant site 42, significant differences occurred between groups Control and PS ($p<0.0001$), PS and ICam ($p<0.0001$), PS and PIC ($p<0.0001$), as well as ICam and PIC ($p=0.0048$). At implant site 44, significant differences were found between the groups Control and PS ($p<0.0001$), PS and ICam ($p<0.0001$), Control and PIC ($p<0.0001$), as well as ICam and PIC ($p<0.0001$) (Fig 9).

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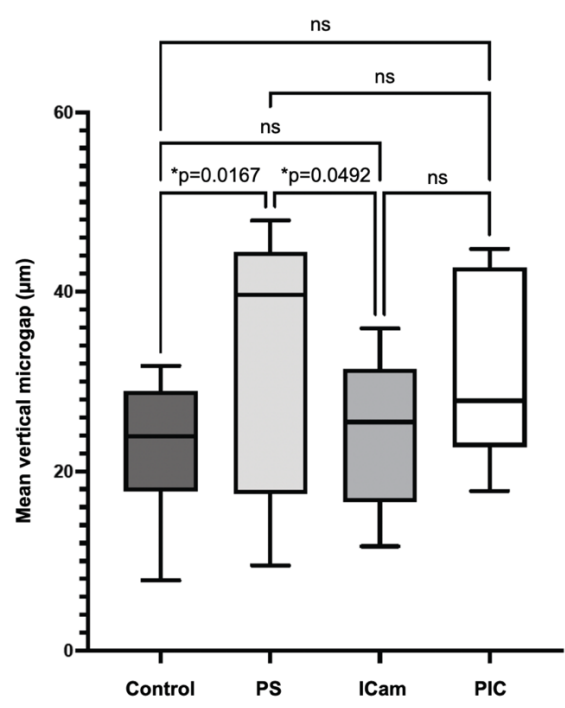


Fig 7 Comparison of vertical microgaps between groups Control, PS, ICam, and PIC.

* indicates significant difference; ns: not significant.

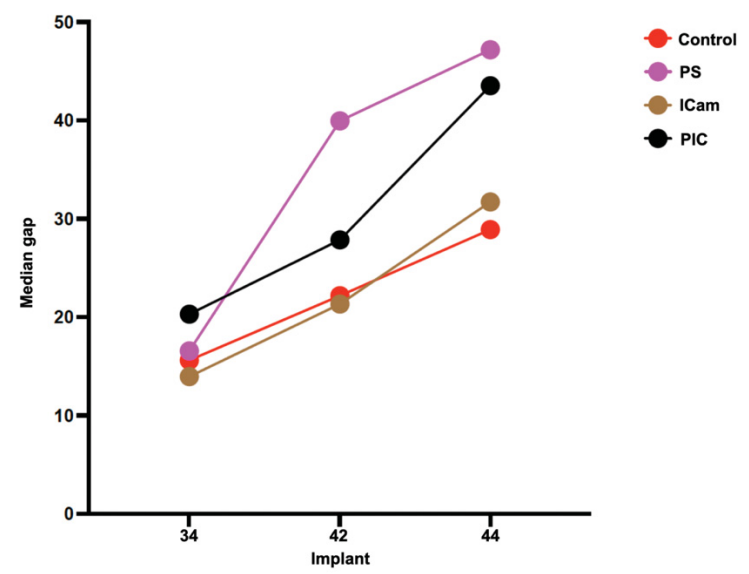


Fig 8 Median vertical microgaps (µm) by implant sites (34, 42, and 44) displayed for groups Control, PS, ICam, and PIC.

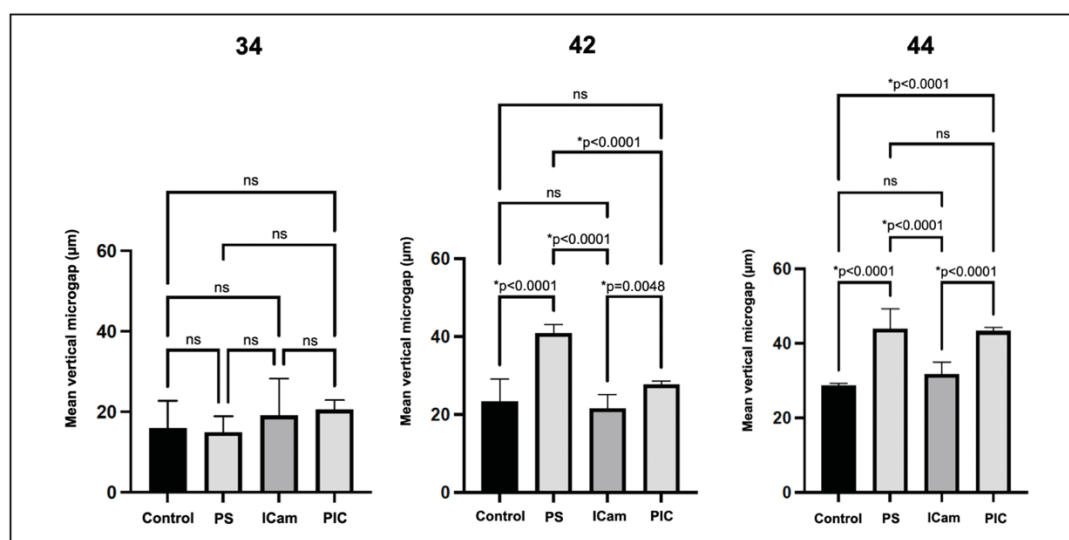


Fig 9 Analysis of vertical microgaps at implant sides 34, 42, and 44. * indicates significant difference; ns: not significant.

Discussion

The present study was designed to evaluate the vertical microgap of milled titanium ISF obtained from digital impressions of one IOS and two PG systems. The null-hypothesis was rejected, as statistically significant differences in vertical microgaps were detected among frameworks of different groups. When scanning edentulous arches, stitching errors occur more frequently due to the limited number of mucosa characteristics.^{6,27} These errors can accumulate and lead to larger errors in subsequent workflow stages, potentially resulting in a lack of framework passivity.^{4,28} Unlike IOS, PG systems are less influenced by external and patient-related factors, such as saliva, ambient light illuminance conditions, span of the edentulous ridge, as well as number, position and angulations of implants.^{29,30} Therefore, the accuracy of the impression taking step may be superior to that of IOS, resulting in a lower cumulative error throughout the entire workflow. The results of the present study support this assumption; however, a significant difference was only detected for one PG device (between ICam and PS).

In dental literature, no consensus exists regarding a threshold of clinical acceptable misfit values.⁴ Most frequently, values of framework misfit were referred to an acceptable threshold of 150µm.^{31,32} In the present study all vertical microgap values were within these threshold. The present findings are in agreement with the majority of previous studies, which also reported that PG systems were more accurate than IOS and conventional impressions for the acquisition of edentulous arches with multiple implants.^{1,9,16,20,33} However, these studies differ from the present one, since they measured the accuracy of digital impressions by comparing RMS-values^{1,16,20,33,34} or inter-implant distance deviations^{9,14,29,35,36} using a 3D evaluation software. In contrast, two other previous studies found that PG exhibited the lowest accuracy when compared to IOS and conventional impression techniques.^{29,35} In these studies, two PG systems (ICam 4D, Imetric 4D Imaging Srl; PIC Camera, PIC Dental) were compared with a splinted conventional impression technique²⁹ and two different IOSs (iTero Element and TRIOS 3).³⁵ The study authors did not provide a rationale for these observed differences.

The majority of studies investigated only one single PG system (PiC Camera^{1,9,13,29,33,36} or ICam 4D^{16,20,35}). One in vitro study was identified that compared both PG cameras.³⁴ Similar to the present findings, no statistically significant differences were observed between PG groups. Furthermore, also the ICam achieved slightly better RMS values than the PIC camera (mean vertical micro gap; ICam: 24.10±7.991 µm, PIC: 30.52± 9.809 µm (present study); mean RMS-values (±CI): ICam: 7.01 µm (6.11; 7.91), PIC: 5.18 µm (4.6, 5.76)³⁴). The accuracy difference between ICam and PIC camera may be related to different technologies, calibration, data processing, and sensor quality. Differences in the material of the optical markers may have also impacted accuracy. While the ICamBody features a metal base, the PIC transfer is designed as a plastic flag. Therefore, it appears to be more susceptible to compressive forces and lacks radiopacity, which may affect its clinical performance.

Further research is needed to explore accuracy variations of PG systems in dental applications.

In the present study vertical microgaps were the highest at implant site,⁴⁴ which is the opposite side and the interface farthest from the point of fixation. This location is the most sensitive for detecting distortions or misfit. However, all misfit values were within a clinically acceptable threshold (most frequently referred to 150µm in dental literature).^{31,32} A previous clinical study also observed higher angular and positional deviations towards distal scan bodies compared to mesial ones.³⁷ Additionally, previous in vitro studies frequently observed higher deviations in posterior areas of digital complete arch implant impressions, which may be related to an error accumulation during the scan across the arch.^{38,39} This theory, however, cannot explain the higher deviations of the PIC camera, since PG is not susceptible to image-stitching errors. A recently published in vitro study assessed the effect of selecting the implant reference on the accuracy of the PIC camera for the acquisition of complete arch implant scans.¹³ For maximizing the accuracy of digital PG impressions, the authors recommended to use the implant placed on the right and most posterior tooth location as reference.

Limitations of the present study include the in vitro conditions and study methodology (evaluation of entire workflow), which may reduce the comparability to other studies. However, evaluating the entire workflow also represents a strength of this study, as it provides the final result of error accumulation in the digital workflow. Variations in milling accuracy, material properties, and potential distortions during manufacturing may have impacted the microgap measurements. A larger sample size could enhance generalizability. Further preclinical and clinical research is necessary to evaluate the accuracy and performance differences among individual PG systems. Additionally, future studies should

investigate the cost-effectiveness, feasibility across diverse clinical environments, and the level of operator training required to consistently achieve high accuracy.

Conclusions

Photogrammetry-based digital impressions resulted in frameworks with lower vertical microgaps compared to those obtained using an intraoral scanner. No statistically significant difference was found between the two photogrammetry systems. The findings highlight photogrammetry as a reliable alternative for implant position recording, offering a more predictable and precise digital workflow for the fabrication of complete arch implant supported frameworks.

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Conflicts of Interest: The authors declare no conflict of interest.

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Monolithic zirconia crowns: effect of thickness reduction on fatigue behavior and failure load

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PURPOSE. The objective of this study was to evaluate the effect of thickness reduction and fatigue on the failure load of monolithic zirconia crowns.

MATERIALS AND METHODS. 140 CAD-CAM fabricated crowns (3Y-TZP, inCorisTZI, Dentsply-Sirona) with different ceramic thicknesses (2.0, 1.5, 1.0, 0.8, 0.5 mm, respectively, named G2, G1.5, G1, G0.8, and G0.5) were investigated. Dies of a mandibular first molar were made of composite resin. The zirconia crowns were luted with a resin composite cement (RelyX Unicem 2 Automix, 3M ESPE). Half of the specimens (n = 14 per group) were mouth-motion-fatigued (1.2 million cycles, 1.6 Hz, 200 N/5 - 55°C, groups named G2-F, G1.5-F, G1-F, G0.8-F, and G0.5-F). Single-load to failure was performed using a universal testing-machine. Fracture modes were analyzed. Data were statistically analyzed using a Weibull 2-parameter distribution (90% CI) to determine the characteristic strength and Weibull modulus differences among the groups. **RESULTS.** Three crowns (21%) of G0.8 and five crowns (36%) of G0.5 showed cracks after fatigue. Characteristic strength was the highest for G2, followed by G1.5. Intermediate values were observed for G1 and G1-F, followed by significantly lower values for G0.8, G0.8-F, and G0.5, and the lowest for G0.5-F. Weibull modulus was the lowest for G0.8, intermediate for G0.8-F and G0.5, and significantly higher for the remaining groups. Fatigue only affected G0.5-F. **CONCLUSION.** Reduced crown thickness lead to reduced characteristic strength, even under failure loads that exceed physiological chewing forces. Fatigue significantly reduced the failure load of 0.5 mm monolithic 3Y-TZP crowns. [J Adv Prosthodont 2021;13:269-80]

KEYWORDS

Ceramics; Zirconium; Monolithic crowns; Fatigue; Computer-aided design

INTRODUCTION

Nowadays, the fabrication of single crowns is the most common restorative procedure in the U.S.¹ According to a current report, the global market of

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dental crowns and bridges will further increase at a compound annual growth rate (CAGR) of 7.78% to USD 3.8 billion in 2026.² In recent years, monolithic restorations have gained popularity due to a reduction of technical complications, such as chipping fractures and a lower price, due to a simple fabrication process compared to layered restorations.³⁻⁵ At this, zirconia and lithium disilicate have become the materials of choice for monolithic applications.⁶ With recent developments in increasing the translucency of zirconia, its indications expanded to the anterior dentition.⁷ As the strongest of all dental ceramics,⁸ zirconia is recommended in stress-bearing areas and allows for non-adhesive cementation.⁶ Literature shows that monolithic zirconia crowns exhibit considerably higher fracture loads than veneered counterparts and lithium disilicate crown restorations.⁹⁻¹¹ For veneered zirconia single crowns, a decreased success rate between 5 and 10 years has been reported.^{4,12} In accordance with previous investigations,^{13,14} chipping of the veneering ceramic was mentioned to be the most frequent clinical complication.⁴ One explanation may be related to residual stress within the veneer layer after the cooling procedure, resulting from the low thermal diffusivity of the Y-TZP framework.^{15,16} Furthermore, a missing cusp supporting anatomical design and parafunctional occlusal forces were mentioned as possible reasons.¹² In terms of survival rates, no statistically significant differences between veneered zirconia and metal-ceramic restorations could be observed.^{4,17} However, significant biological complications following more invasive preparation techniques, especially the loss of abutment tooth vitality, were reported for metal-ceramic crowns.⁴ Due to a superior fracture strength of zirconia,^{9,11,18,19} monolithic restorations can be fabricated with reduced layer thicknesses and thus potentially prevent biological complications. Yet, no long-term studies on monolithic zirconia restorations are available. A 3-year short-term study²⁰ recorded survival-rates of 100% for monolithic zirconia single crowns with a minimum occlusal thickness of 0.5 mm. Other studies reported survival rates of 91.5% after 3.5 years²¹ and 98% after 5 years²² for monolithic zirconia crowns with an occlusal thickness between 0.5 and 1 mm. Whereas some literature is available on reduced

thicknesses of 3Y-TZP ceramics,²³⁻²⁶ the effect of fatigue has still not been extensively addressed.

Zirconia materials appear susceptible to bulk fracture in thin areas.^{24,26} Consequently, there is still no consensus on how thin restorations can be fabricated,²⁵ and the impact of thermomechanical fatigue on the mechanical properties of thin monolithic zirconia crowns warrants further research. Therefore, the aim of the present study was to evaluate the effect of thickness reduction on the failure load and 5-year *in-vitro* survival rate of monolithic 3Y-TZP zirconia crowns. The tested null hypotheses were that (i) ceramic layer thickness and (ii) fatigue application do not influence the failure load of posterior 3Y-TZP monolithic zirconia crowns.

MATERIALS AND METHODS

In this *in-vitro* study a total number of 112 crowns were divided into four groups ($n = 28$ per group) with different ceramic layer thicknesses (1.5 mm (G1.5), 1.0 mm (G1), 0.8 mm (G0.8), 0.5 mm (G0.5)) (Fig. 1). 28 crowns with a 2.0 mm (G2) ceramic thickness served as control. This control group was already investigated in an earlier study.²⁷ Monolithic crowns of a first mandibular molar (tooth 46 FDI) were fabricated out of a 3Y-TZP zirconia ceramic with a flexural strength of > 900 MPa (inCorisTZI, Dentsply Sirona, Bensheim, Germany).²⁸ All zirconia crowns were designed with a CAD-CAM software (Cerec InLab 4.0, Dentsply Sirona) and a multilayer design was chosen in order to create a separate abutment die and a monolithic crown restoration. By changing the configuration parameters (Fig. 2), the dimension of the die was adjusted depending on the layer thickness of the respective crown, while the occlusal morphology of the crown restoration remained identical. After milling, (inLab MC X5, Dentsply Sirona) crowns were dyed (inCoris TZI Coloring Liquids, Dentsply Sirona) and sintered (inFire HTC speed, Dentsply Sirona) according to manufacturer's recommendations. A conventional sintering procedure was used at 1510°C with a dwell time of 120 minutes.²⁸ The initial heating rate was 25°C per minute until a temperature of 800°C was reached. Afterwards, the furnace was heated from 800°C to 1510°C with a heating rate of 15°C per min-

	Control	Test							
Restorative Groups	inCoris TZI G2 n = 28	inCoris TZI G1.5 n = 28	inCoris TZI G1 n = 28	inCoris TZI G0.8 n = 28	inCoris TZI G0.5 n = 28				
Crown Design	monolithic								
Cement	RelyX Unicem 2 Automix								
Fatigue	No fatigue	Fatigue (1.2 million / 200 N)	No fatigue	Fatigue (1.2 million / 200 N)	No fatigue	Fatigue (1.2 million / 200 N)	No fatigue	Fatigue (1.2 million / 200 N)	No fatigue
Failure Load	Single load to failure (universal testing machine)								

Fig. 1. Test set-up.

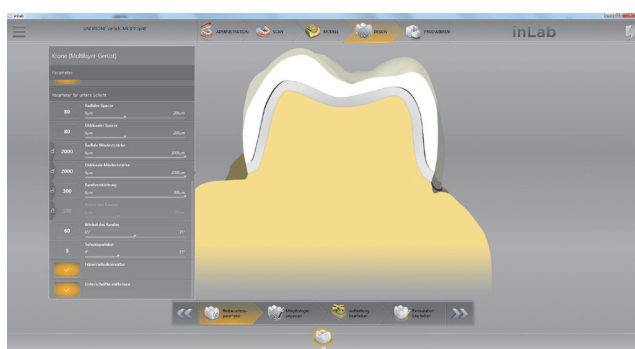


Fig. 2. Definition of the different parameters. Yellow = die, grey = varying enlargement of the die, white = monolithic crown (Screenshot, provided by Dentsply Sirona).

ute. For slow cooling, the furnace was programmed to remain closed up to a temperature of 200°C using a cooling rate of 30°C per minute. The glaze firing was carried out with a closing time of 4 minutes and a starting temperature of 400°C, followed by a temperature increase of 45°C per minute until the final temperature of 870°C was reached. This temperature was maintained for 1 minute, then cooled down to 600°C until the furnace was opened.

112 composite resin master dies were made using a dentin-analog light-curing nanohybrid composite (Tetric Evo Ceram A2 Dentin, Ivoclar Vivadent AG, Schaan, FL; elastic modulus of 8.6 GPa, information provided by manufacturer). Therefore, the dies were duplicated and a plaster replica die was generated for each monolithic crown. Impressions of the plaster dies were made using a polyvinylsiloxane impression material (Affinis, Coltene AG, Altstätten, Switzerland). Subsequently, the negative form was filled up with 1.5 mm thick nanohybrid composite layers (Tetric Evo Ceram) and was light-cured (Polofil Lux, Voco, Cuxhaven, Germany) with approx. 780 mW/cm² from all directions for 20 seconds.

The resin dies were stored in distilled water in an incubator (IN30, Memmert GmbH+Co KG, Schwabach, Germany) at 37°C for a minimum of 30 days (mean: 31 days; min: 30 days; max 32 days). Afterwards, specimens were embedded in a cold polymerizing polyester resin (Technovit 4000, Heraeus Kulzer, Hanau, Germany). Resin dies were pretreated with pumice powder (Picodent, Wipperfurth, Germany), rinsed with water, dried, and cleaned with 60% ethanol. Prior to cementation, a layer of MDP-containing prim-

er (Clearfil ceramic primer Plus, Kuraray Noritake, Tokyo, Japan) was applied to the internal surface of the zirconia crowns (no air-particle abrasion was employed). All crowns were then cemented with a resin composite cement (RelyX Unicem 2 Automix, 3M ESPE, Seefeld, Germany) according to manufacturer's instructions. Excess cement was carefully removed with a scaler after light-curing of 2 seconds (Polofil Lux, Voco, Cuxhaven, Germany). Afterwards, each surface was light-cured for another 20 seconds at around 780 mW/cm². Finally, all specimens were stored in distilled water for a minimum of 7 days (mean: 8 days; min: 7 days; max 8 days) to allow hydration of the resin cement.

Half of the specimens (G2-F, G1.5-F, G1-F, G0.8-F, G0.5-F) of each group were exposed to mouth-motion fatigue (-F) using a chewing-simulator (CS-4.8, SD Mechatronik, Feldkirchen-Westerham, Germany) with a load of 200 N at 1.6 Hz for 1.2 Mio cycles and simultaneous thermocycling (5°C to 55°C, dwell time 120 s). This corresponds to a clinical ageing time of 5 years.²⁹ Steatite spheres (Hoechst Ceram Tec, Wunsiedel, Germany) with a diameter of 6 mm served as antagonists. To simulate a lateral masticatory movement, the steatite spheres moved 0.5 mm horizontally from the distolingual cusp towards the central fissure. Subsequently, all crowns of each group underwent single load to failure (SLF) testing in a universal testing machine (Zwick Z010/TN2S, ZwickRoell GmbH & Co KG, Ulm, Germany). The load was applied axially with a metal sphere ($r = 3.18$ mm) towards the central fossa. The force was applied parallel to the longitudinal axis of the specimen at a test speed of 1.5 mm/min. Failure loads were recorded with a computer software (testXpert II V7.1, ZwickRoell) in Newton [N] for statistical analysis. Finally, all specimens were visually evaluated for failure analysis using a stereo light microscope (SZH 10, Olympus, Hamburg, Germany).

Reliability demonstration test (RDT tool, Weibull++ v.21, Reliasoft, Tucson, AZ, USA) was used to determine sample size, considering mean time to failure of 3000 N (estimated average load to failure range for groups between 0.5 mm to 2 mm thicknesses), under 90% confidence bounds, using Weibull 2-parameter distribution, assuming Beta value = 5 (estimated average Beta), resulting in a sample size of 11. To account

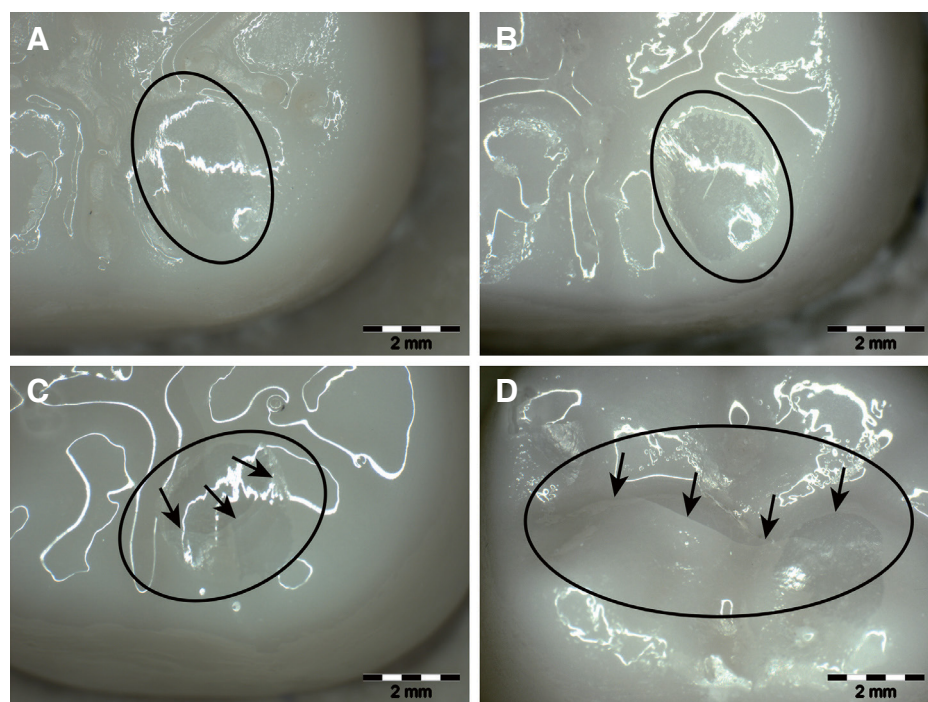
for potential fractures during fatigue, 3 additional samples per group were used, totaling 14 samples per group. Data were statistically analyzed using a Weibull 2-parameter distribution (Weibull++ v.21, Reliasoft, Tucson, AZ, USA) (90% CI) to determine the characteristic strength (N) and Weibull modulus (m) differences among groups. Differences among groups were also depicted in a contour plot (characteristic strength vs. Weibull modulus). The probability of survival (reliability) was also plotted to show group survival distribution as a function of load to failure.

RESULTS

All crowns of G2, G1.5, and G1 presented no failures after 1.2 million fatigue cycles, which corresponds to a clinical 5-year survival.^{29,30} Three crowns (21%) of G0.8 and five crowns (36%) of G0.5 showed cracks after fatigue. Due to the limited extension of these cracks, all crowns were still exposed to single load to failure (SLF). The observed cracks of G0.8-F and G0.5-F were located in the area of the distolingual cusp, which corresponds with the lateral sliding movement from the distolingual cusp to the central fossa (Fig. 3). After fatigue exposure, all crowns showed superficial wear facets of the glazing material at the distolingual cusp (Fig. 3).

The characteristic strength and Weibull modulus results with differences among groups, based on overlap between upper and lower confidence intervals, are presented in Table 1. Characteristic strength (N) was the highest for G2 compared to G1.5 with significant differences between them. Although decreased characteristic strength values were observed after fatigue in G2-F and G1.5-F, they were not significantly different when compared with their respective non-fatigued counterparts. Intermediate values were observed for G1 and G1-F, followed by significantly lower values for G0.8, G0.8-F, and G0.5, and the lowest for G0.5-F. Weibull modulus was the lowest for G0.8, intermediate for G0.8-F and G0.5, and significantly higher for the remaining groups. Different superscript letters indicate significant differences among groups. Fatigue did not have a statistically significant influence, except for G0.5 (Table 1). A contour plot (Fig. 4A) is presented as an instructive graphical method

Fig. 3. Monolithic zirconia crowns with different layer thicknesses after loading ($1.5 \times$ magnification). The circle shows the attrition caused by the antagonists during fatigue. The arrows mark the cracks. (A) G1.5-F, (B) G1-F, (C) G0.8-F, (D) G0.5-F. Cracks were found in groups 0.8-F and 0.5-F.



to depict characteristic strength vs. Weibull modulus significant differences between datasets (no overlap of confidence bounds). Each contoured region represents possible values given both parameter's combination, and difference at 90% level is detected if contour overlap between groups does not exist.³¹ Figure 4B depicts the probability of survival (reliability) distribution as a function of characteristic strength.

After SLF, all crowns showed catastrophic bulk fractures. Crowns fractured mostly into 2 - 5 main fragments that could be repositioned. No differences

could be observed between the individual groups. However, failure analysis revealed that cracks propagated from the occlusal loading point to the cervical crown margin (Fig. 5). The crack origin was mainly located in the distolingual region of the occlusal surface initiating from the area of loading. Fractures within the die material only occurred in groups with a layer thickness of 1.0 mm (G1:0%, G1-F:7.14%), 1.5 mm (G1.5:42.85%, G1.5-F:28.57%) and 2.0 mm (G2:100%, G2-F:100%).

Table 1. Characteristic strength (N) and Weibull modulus (m) for the different groups. Different superscript letters indicate significant differences

	G2	G2-F	G1.5	G1.5-F	G1	G1-F	G0.8	G0.8-F	G0.5	G0.5-F
upper limit	6389.9	5895.6	5244.9	4911.7	2969.7	2972.89	1827.4	1691.66	1609.35	1170.87
Characteristic strength (N)	5934.2 ^A	5540.2 ^{A,B}	5008.3 ^{B,C}	4576.6 ^C	2783.4 ^D	2785.16 ^D	1555.18 ^E	1533.08 ^E	1454.88 ^E	1121.81 ^F
lower limit	5510.9	5206.3	4782.3	4264.3	2608.7	2609.27	1323.5	1389.36	1315.23	1074.81
upper limit	8.7	10.3	14.1	9	10	9.84	4	6.79	6.48	15.6
Weibull modulus (m)	6.3 ^{A,B}	7.5 ^{A,B}	10 ^A	6.5 ^{A,B}	7.2 ^{A,B}	7.15 ^{A,B}	2.89 ^C	4.68 ^B	4.59 ^B	10.7 ^A
lower limit	4.5	5.4	7.1	4.8	5.17	5.19	2	3.22	3.26	7.39

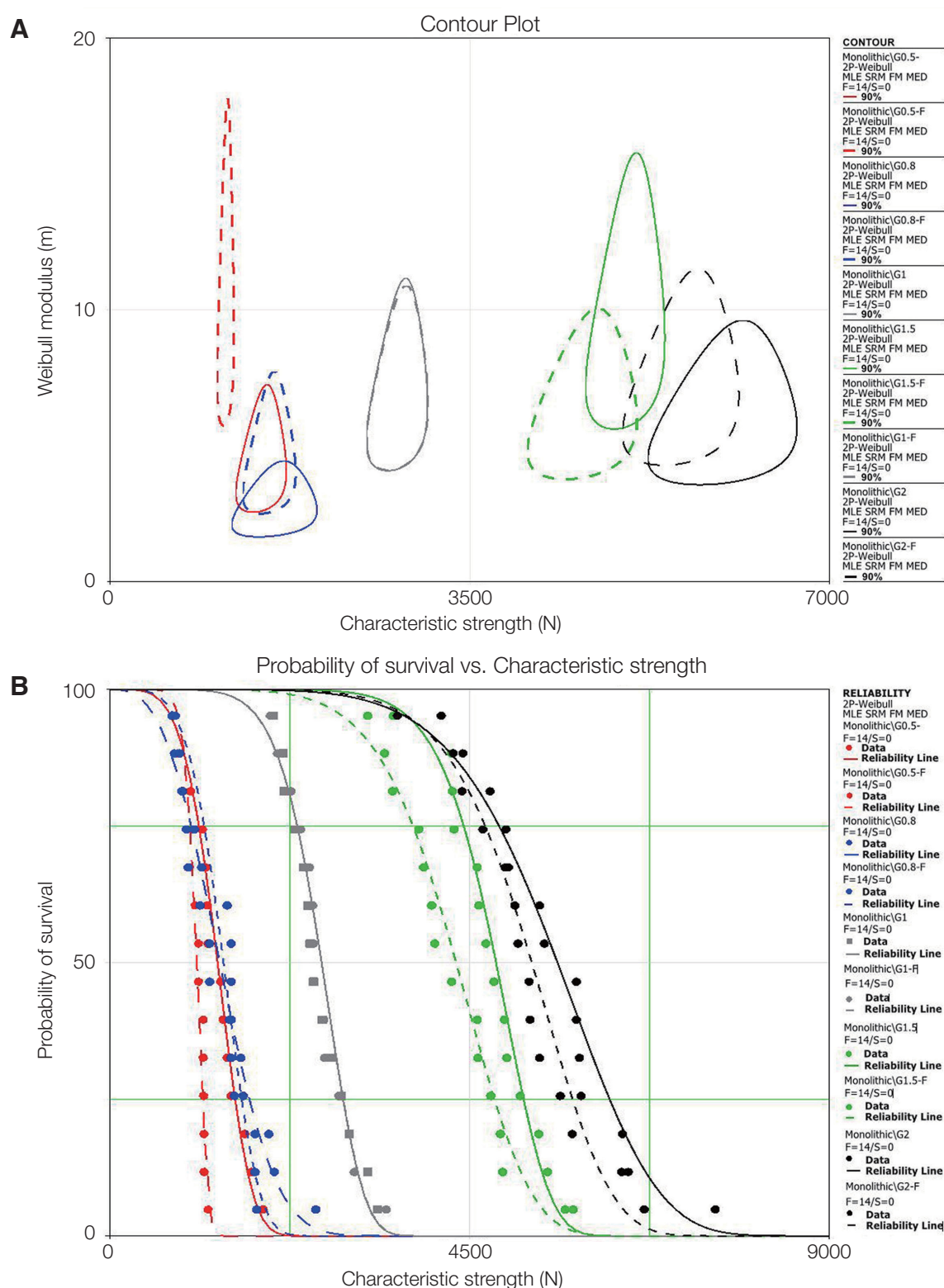
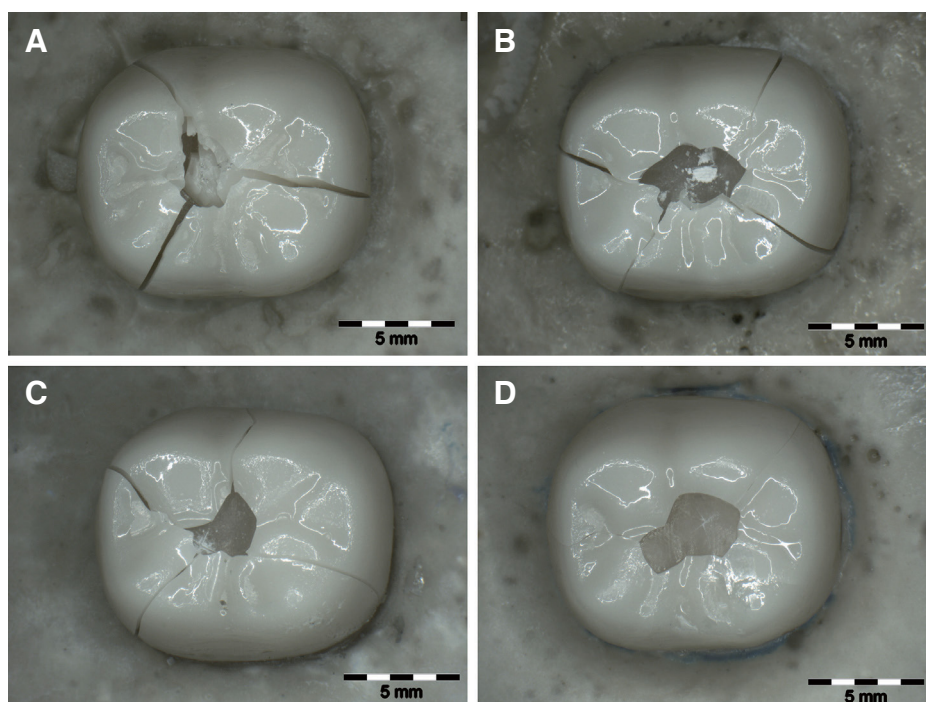


Fig. 4. (A) Two-parameter Weibull contour plot (Weibull modulus [m]) vs. characteristic strength [N] for G2 (black), G1.5 (green) G1.0 (grey), G0.8 (blue) and G0.5 (red) in which solid contour lines represent non-fatigued, and dotted lines represent groups subjected to fatigue prior to single load to failure test. The only group showing no overlap between contours before and after fatigue is G0.5, meaning that fatigue significantly decreased its characteristic strength. (B) Probability of survival vs. characteristic strength shows a general trend in crown survival increase for increased thickness. G1 and G1-F (grey solid and dotted reliability lines, respectively) present an overlap due to similar sample failure distribution.

Fig. 5. Monolithic zirconia crowns after single load to failure ($1.5 \times$ magnification). (A) G1.5, (B) G1, (C) G0.8, (D) G0.5.



DISCUSSION

The present study showed that all tested crowns withstood failure loads above physiological chewing forces (50 - 250 N physiological, 500 - 900 N para-functional).^{26,32} Nevertheless, cracks during fatigue were observed in G0.8 and G0.5. Therefore, the tested null hypothesis was rejected as layer thickness and fatigue application (G0.5) did influence the failure load of posterior 3Y-TZP monolithic zirconia crowns. This can be explained due to the fact that five of the crowns (36%) with fatigue exposure showed cracks after chewing simulation causing a reduction in fracture resistance. It was observed that when the layer thickness was doubled from 0.5 mm to 1.0 mm and from 0.8 mm to 1.5 mm, characteristic strength values increased two- to threefold. Similar observations were made in other *in vitro* studies.^{11,26}

Considering the significant decrease in characteristic strength observed for crowns of 0.8 mm compared to those of higher thickness and especially evident for the 0.5 mm group subjected to fatigue, it may be suggested that crown thicknesses ≥ 1 mm, supported by dentin, may be less prone to fracture. The clinical longevity in long-term remains uncertain. A reduced

layer thickness of 1.0 mm seems to be the limit with no negative influence on the 5-year *in vitro* survival rate and should thus be preferred. For this reason, most manufacturers do not recommend to reduce the occlusal layer thickness of monolithic zirconia crowns below 1 mm in posterior areas.^{28,33,34} In contrast, some previous evidence suggests that it can be reduced up to 0.5 mm keeping a sufficient strength to endure maximum chewing forces up to 900 N.^{19,23-25}

Previous studies investigated different layer thicknesses of 0.5 to 1.5 mm with different chewing loads and different numbers of masticatory cycles.^{10,23,26,35} Due to different experimental protocols and test setups, a direct comparison is difficult. Overall, thermomechanical cycling is the most commonly used method for fatigue testing.^{24,26,36} Some other studies only used mechanical cycling¹⁰ or omitted fatigue exposure before failure load testing.^{19,25} A previous study could show that the flexural strength of 3Y-TZP disc-shaped specimens was predominately affected by mechanical or thermomechanical cycling.³⁷ Therefore, specimens were exposed to mechanical cycling (15,000,000 cycles/3.8 Hz/200 N), thermal cycling (6,000 cycles/5 - 55°C/30 s), and thermomechanical cycling (1.2 million cycles/3.8 Hz/200 N and ther-

mal-cycling at a temperature range from 5°C to 55°C for 60 s each). In the present study, half of the specimens were subjected to a similar thermomechanical fatigue protocol (1.2 million cycles/1.6 Hz/ 200 N and thermal-cycling at temperature range from 5°C to 55°C for 60 s each).

Reported loads during clinical mastication vary considerably.^{32,38} As previously described, force peaks of 200 N in the anterior region, 350 N in the posterior, and 1000 N with bruxism occur.³⁸ Many previous investigations used average chewing forces that were considerably lower (between 49 - 80 N) compared to the loads applied in the present test set-up.^{10,24,39} A recent study determined the test conditions in a preliminary pilot study and finally used a force of 200 N for preloading in a fatigue test set-up.³⁶ In the present study, a chewing force of 200 N was also chosen, since it considers a certain safety load and seems to avoid either too low and too high loads.^{32,36}

Characteristic strength values of this present study are similar to values detected in other studies: In a study investigating different layer thicknesses of monolithic 3Y-TZP zirconia crowns (Lava Frame, 3M ESPE), the following mean fracture load values were measured: 4109 ± 610 N (1.5 mm), 3068 ± 233 N (1.2 mm), 2429 ± 315 N (1.0 mm), 1814 ± 68 N (0.8 mm) and 1308 ± 111 N (0.6 mm).¹¹ Comparing 3Y-TZP zirconia with different alumina contents, fracture load values varied from 450 to 3248 N (Lava Zirconia, 3M ESPE) and 438 to 3487 N for 3Y-TZP with a lower alumina content of 0.1% (Lava Plus, 3M ESPE).²⁶ However, lower values of 691 - 2048 N for monolithic zirconia crowns with layer thicknesses of 2.0, 1.5, 1.0 and 0.5 mm²⁵ and substantially higher values of 5558 ± 522 N (Lava Plus Zirconia, 3M ESPE) for crowns with an occlusal thickness of 0.5 mm¹⁹ were found in the literature. However, specimens were partly unexposed to fatigue.^{19,25} Since also the experimental set-up differs, a direct comparison is not possible. Another previous study compared monolithic 3Y-TZP zirconia crowns (Zikon Biostar HT, Siladent Dr. Böhme & Schöps GmbH, Goslar, Germany) with occlusal layer thicknesses of 0.5 and 0.2 mm and found similar failure load values for 0.5 mm thin crowns (1628 ± 174 N adhesively bonded, 1357 ± 340 N cemented).²⁴

According to manufacturer's recommendations, an

adhesive cementation for 3Y-TZP zirconia crowns is not required, when a sufficient mechanical retention and crown thickness is given.^{28,40} However, composite resin cements or ceramic primer with special adhesive monomers, like MDP, are recommended for the bonding to zirconia.⁴¹ For this reason, an MDP-containing universal primer (Clearfil ceramic primer Plus, Kuraray Noritake) and a self-adhesive dual-curing composite luting cement (RelyX Unicem 2 Automix, 3M ESPE) were used in this study. But, cementation procedures are controversially discussed in the present literature.^{40,42}

As a limitation of this *in vitro* study, it has to be stated that dentin-analog resin dies instead of extracted human molars were used, which do not exactly represent the structure of a real tooth. On the other hand, extracted teeth are difficult to standardize due to age and size difference.⁴³ The applied composite resin reveals a similar elastic modulus (Tetric Evoceram: 8.6 GPa) compared to human dentin (10 - 16 GPa), and was also applied in comparable studies.^{15,19,26,44,45} A thermomechanical cycling protocol followed by a static test (SLF) was chosen. This commonly used method^{24,26,46} obtains reliable information and allows ranking of all-ceramic systems, according to given variables such as thicknesses.

Clinical failures are usually based on slow crack growth mechanisms,⁴⁷ which lead to a continued reduction of initial strength over the years.⁴⁸ Two important crack mechanisms for bulk fractures are reported: firstly, radial cracks caused by tensile stress propagating from the intaglio to the occlusal surface and secondly, cracks caused by contact stress at the occlusal surface propagating downwards.⁴⁹ After SLF, the crack origins were mainly located at the distolingual area of the occlusal surface. This can be attributed to a change of the ceramic structure after fatigue exposure caused by the sliding movements at the distolingual cusp to the central fossa. A recently published clinical study found a grain transformation in the occlusal loading areas of monolithic zirconia crowns after two years in situ.⁵⁰ In the present study, all crowns showed catastrophic bulk fractures after SLF. The observed cracks propagated from the occlusal loading point to the cervical crown margins. This fracture pattern initiating from the occlusal loading

points has also been reported in clinical^{12,50} and *in vitro* studies.^{11,25,51}

In terms of failure mode, no differences could be observed among the groups of this study. Interestingly, all dies of G0.5 and G0.8 remained intact, while the dies of G2, G1.5, and G1 showed fractures. In these groups, partial and longitudinal fractures occurred and a few dies even showed bulk/catastrophic failures. The simulated abutment teeth/ dies of G0.8 and G0.5 were thicker than the dies of G2, G1.5, and G1. This might explain why no die fractures occurred in G0.8 and G0.5. Furthermore, it should be considered that in the present experimental set-up, stiff zirconia specimens were loaded by using a force of 200 N. According to a previous investigation,⁵² the velocity of the weight and the resulting impulse depend on the stiffness of the specimen material. For stiff specimens, several impacts /‘bouncing’ were detected leading to substantial measurement errors. Therefore, low impact velocities are recommended, especially for stiff specimen loading, and were applied in the present study.⁵² Axial loading might produce several uncontrolled contact points.⁵³ Therefore, other studies used a tin foil distributing the force equally to the cusps.^{9,11,19,39} Another investigation applied an inverse V-shaped two-plane intender for achieving multiple contacts more equally and in order to reproduce more clinically relevant fractures.⁵⁴ Some authors suggest that an angulation of the specimens could better represent clinical scenarios, due to the more complex stress state caused by sliding actions.^{9,26} Other authors used an off-axis-loading of 30° for the investigation of flat specimens in order to simulate cusp inclination.^{55,56} However, in the present study, a sliding component was achieved by the lateral movement of the steatite sphere during fatigue testing.

In recent years, increased translucency of zirconia could be achieved by increasing the yttria content or the cubic phase (4Y-PSZ and 5Y-PSZ).⁷ Literature shows that these translucent zirconia materials are associated with lower mechanical properties because phase transformation toughening is absent due to predominance of cubic phase, which makes them more prone to failure.^{36,57-59} However, it is expected that in the future, zirconia materials will be further enhanced, regarding translucency, fluorescence, and

strength. Further investigations on failure load, pre-treatments, and thermomechanical longevity of 4Y- and 5Y-TZP crowns should be addressed in the future.

CONCLUSION

Within the limitations of this study, the following can be concluded: layer thickness has a statistically significant influence on the characteristic strength of 3Y-TZP monolithic zirconia crowns. All crowns with a layer thickness above 1.0 mm withstood a simulated 5-year clinical application without fracture. Yet, all groups achieved characteristic strength values that exceed physiological chewing forces. A preparation design resulting in a reduction for a ceramic layer thickness of 1.0 mm can be recommended. Further laboratory and clinical studies are necessary to confirm these results.

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RESEARCH ARTICLE

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Minimally invasive CAD/CAM lithium disilicate partial-coverage restorations show superior in-vitro fatigue performance than single crowns

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Abstract

Objective: To analyze the influence of restoration design (partial-coverage restoration vs. crown) and ceramic layer thickness on the performance and failure loads of CAD/CAM-fabricated lithium disilicate (LDS) reconstructions on molars after fatigue.

Materials and Methods: Seventy-two posterior monolithic CAD/CAM-fabricated LDS restorations (IPS e.max CAD, Ivoclar Vivadent) with different occlusal/buccal ceramic layer thicknesses (1.5/0.8, 1.0/0.6, and 0.5/0.4 mm) and restoration designs (PCR: non-retentive full-veneer/partial-coverage restoration, C: crown,) were investigated and divided into six groups ($n = 12$, test: PCR-1.5, PCR-1.0, PCR-0.5; control: C-1.5, C-1.0, C-0.5). LDS restorations were adhesively bonded (Variolink Esthetic DC, Ivoclar Vivadent) to dentin-analogue composite dies (Z100, 3M ESPE). All specimens were subjected to thermomechanical loading (1.2 million cycles, 49 N, 1.6 Hz, 5–55°C) and exposed to single load to failure testing. Failure analysis was performed with light and scanning electron microscopies. Data were statistically analyzed using ANOVA, Tukey-Test, and t -test ($p < 0.05$).

Results: Eight crown samples (C-0.5) and one PCR specimen (PCR-0.5) revealed cracks after fatigue, resulting in an overall success rate of 87.5% (crowns: 75%, PCRs: 96.88%). Direct comparisons of PCRs versus crowns for thicknesses of 0.5 mm ($p < 0.001$) and 1.0 mm ($p = 0.004$) were significant and in favor of PCRs. Minimally invasive PCRs (0.5 and 1.0 mm) outperformed crowns with the identical ceramic

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thickness. No difference was detected ($p = 0.276$) between thickness 1.5 mm PCR and crowns.

Conclusions: Minimally invasive monolithic CAD/CAM-fabricated posterior LDS PCRs (0.5 and 1.0 mm) resulted in superior failure load values compared to minimally invasive crowns. Minimally invasive crowns (0.5 mm) are prone to cracks after fatigue.

Clinical Significance: Minimally invasive CAD/CAM-fabricated LDS PCR restorations with a non-retentive preparation design should be considered over single crowns for molar rehabilitation.

KEYWORDS

ceramics, computer-aided design, fatigue, full-veneer, partial-coverage restoration

1 | INTRODUCTION

Nowadays, the most frequent reconstructive treatment in the United States is still the single-tooth restoration with annual growth rates of up to 8% for single crowns worldwide.^{1–3} In times of digital dentistry, the steadily evolving Computer-Aided Design (CAD)/Computer-Aided Manufacturing (CAM) technology leads to a trend toward monolithic restorations in all material classes.^{4–7} Monolithic reconstructions fulfill the potential to avoid chipping and delamination problems, which are common in veneered bi-layer systems, while reducing tooth substance removal and production costs at the same time.^{8–12}

Increased prevalence of severe tooth wear caused by bio-corrosive defects, such as erosion, abrasion, attrition or combinations of these, especially in young patients, accelerated the development of minimally invasive treatment concepts.^{13,14} To rehabilitate the posterior dentition in a minimal invasive approach, occlusal veneers evolved as the preferred treatment choice.^{15–18} If defects additionally involve buccal and cervical areas, clinicians must choose between a full coverage-crown preparation or an occlusal restoration with an accessory Class V composite filling. Drawbacks of this treatment approach are the increased reduction of sound tooth structure for full-crown preparations and the reduced longevity of Class V fillings in the worn dentition.^{19,20} Partial-coverage restorations (PCR) or so called “full-veneers”, which include occlusal, labial and if necessary proximal areas offer a minimally invasive treatment alternative for such teeth.^{21,22} With the benefit of superior esthetics in patients with a strong visibility of premolars and first molars, these PCRs may serve as an esthetic indirect restoration while preserving as much tooth substance as possible.^{23,24} Moreover, social-network content and video-calls (“zoom-boom”), especially during the COVID-19 pandemic, have increased critical self-perception of one's own smile and outward appearance as well as the wish for esthetic self-improvement of patients.^{25,26} A recent survey by the British Orthodontic Society underlies this: 62% of orthodontists recorded an increase in demand for esthetic treatments during COVID-19.²⁶

Lithium disilicate glass-ceramics (LDS) offer superior esthetics^{27,28} and a reliable clinical long-term survival.^{29–31} The clinical performance of LDS CAD/CAM crowns is high, with survival rates of 93% after 6 years³² and 80.1% after 15 years.³³ CAD/CAM-fabrication of LDS reconstructions provides a standardized manufacturing process and enables an improvement of cost and time effectiveness.³⁴ Nonetheless, the manufacturers' guidelines suggest a minimum restoration thickness of 1.0 mm for posterior crowns and partial-coverage restorations to avoid restoration fractures during clinical service.^{34–37} Compared to traditional crown preparations with a recommended occlusal reduction of 1.5–2 mm, minimally invasive preparation designs require a less invasive tooth preparation.^{38–41} LDS CAD occlusal veneers with an average occlusal reduction of 0.4–0.6 mm in the central groove and 1.0–1.3 mm at the cusp tips presented a survival rate of 100% after 3 years.⁴² High survival rates of 96.49% after 16.9 years were also reported for LDS posterior crowns and partial-coverage restorations (IPS e.max Press) (≥ 1 or $<$ than 1 mm).³¹

Yet, in-vitro literature is controversial if LDS CAD crowns with a minimally invasive layer thickness of 0.7–1 mm are clinically applicable.^{4,39,43–45} The limited evidence of in-vitro studies, which are available on monolithic LDS (IPS e.max CAD and IPS e.max Press) crowns with a reduced layer thickness of 0.5 mm reported a high variability of failure loads (values between 369.2 ± 117.8 and 827 ± 318 N^{34,46,47}). In contrast, minimally invasive LDS occlusal veneers (IPS e.max Press and IPS e.max CAD) showed extraordinary results in a systematic review on in-vitro and clinical studies.^{15,48–50} Nevertheless, clear recommendations for a minimally invasive preparation design for cases in which defects involve occlusal as well as buccal or cervical areas are currently missing. Therefore, the aim of this in-vitro study was to evaluate the fatigue survival and failure load of CAD/CAM fabricated LDS standard (1.5 mm), thin (1.0 mm), and ultrathin (0.5 mm) PCRs compared to crowns on molars after thermomechanical loading. The research hypotheses were that (i) the restoration design (PCR vs. crown) and (ii) ceramic layer thickness would affect the failure load of posterior CAD/CAM LDS restorations.

2 | MATERIALS AND METHODS

In this in-vitro study a total of 72 specimens were randomly divided into two groups according to the type of restoration (Test: full-veneer/partial-coverage restoration (PCR) vs. control: crown (C)) and further subdivided according to occlusal/buccal ceramic layer thickness (0.5/0.4, 1.0/0.6, and 1.5/0.8 mm), resulting in subgroups of $n = 12$ specimens. The test-set up is displayed in Figure 1.

2.1 | Specimen preparation

To standardize test specimens, a maxillary first molar of a typodont model (frasaco-model, frasaco, Tett nang, Germany) was used to simulate a realistic clinical condition. Before preparation, two silicone impressions (TwinDuo, Picodent, Wipperfürth, Germany) were made to control tooth substance removal in a bucco-palatal and mesio-distal direction.

A single experienced prosthodontist performed the master die preparations for the type of restoration designs (PCR vs. crown) and

different ceramic layer thicknesses (0.5, 1.0, and 1.5 mm). In total six master dies were prepared. Preparation was performed with a 4.5 optical enlargement, coarse and fine diamond burs (no. 370314 035, no. 8370314 035, no. 879314 012, and no. 8879314 012, Komet, Lemgo, Germany) at high speed and under constant air-water-spray cooling. The PCR preparation design was non-retentive (no occlusal/proximal box preparation) and involved occlusal, buccal, and proximal areas in the corresponding thickness (Group 0.5: 0.5/0.4 mm; Group 1.0: 1.0/0.6 mm; Group 1.5: 1.5/0.8 mm). To facilitate adhesive cementation and seating of PCRs two diagonal shallow notches (0.2 mm depth) were additionally prepared (Figures 2 and 3). The buccal area received a chamfer finish line in the respective thickness (Figure 3). The full-coverage crown preparation was performed accordingly without notches (Figure 3). Preparation depth was controlled with silicone keys and a periodontal probe.

To generate 72 dentin-analogue abutments, a resin-based composite (Filtek Z100, 3M ESPE, Neuss, Germany) with an elastic modulus of 18 GPa,⁵¹ which is similar to that of human dentin with 16–18 GPa,⁵² was chosen. Impressions of the six master dies were obtained with a polyvinylsiloxane material (Identium, Kettenbach,

Group	LDS IPS e.max CAD n = 72					
	Partial Coverage PCR-0.5 n = 12	Partial Coverage PCR-1.0 n = 12	Partial Coverage PCR-1.5 n = 12	Crown C-0.5 n = 12	Crown C-1.0 n = 12	Crown C-1.5 n = 12
Design	Monolithic					
Cement	Variolink Esthetic DC					
Fatigue	Fatigue (1.2 million @ 49 N)					
Failure Load	Single load to failure (universal testing machine)					

FIGURE 1 Test set-up.

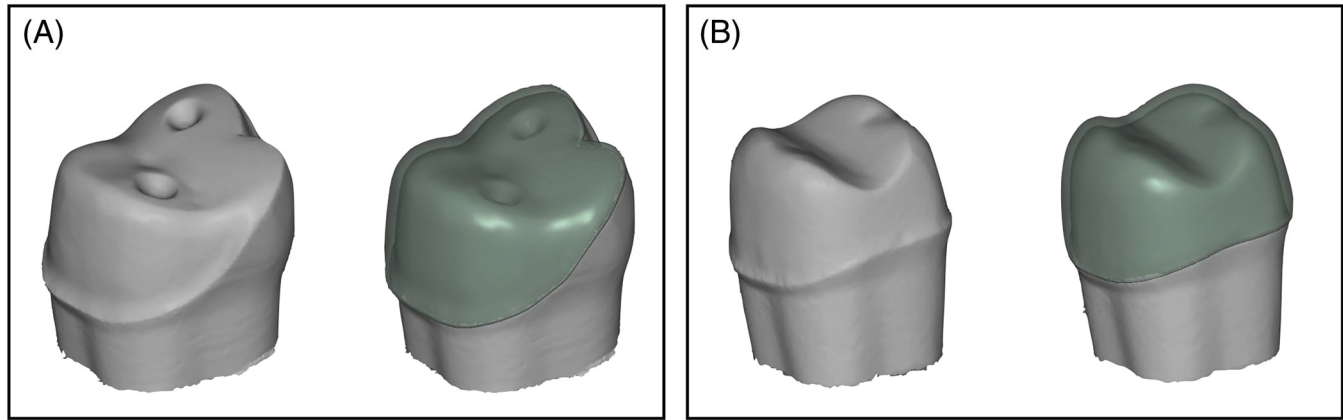


FIGURE 2 Preparation and restoration designs of (A) Full-veneer/Partial-coverage restoration and (B) Complete crown.

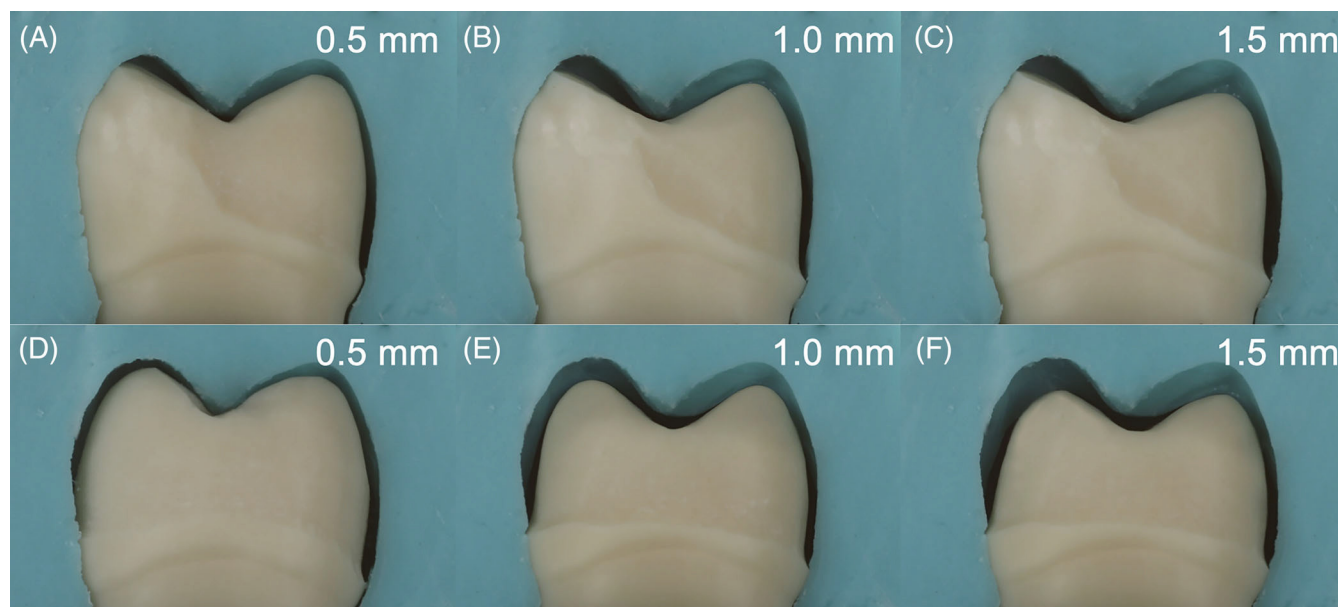


FIGURE 3 Preparation design of a maxillary first molar with respective silicon keys; (A–C) Non-retentive partial-coverage restoration. (A) Group 0.5 (occlusal 0.5 mm/buccal 0.4 mm); (B) Group 1.0, (1.0/0.6 mm) and (C) Group 1.5 (1.5/0.8 mm); (D–F) Complete crown restoration, (D) Group 0.5 (0.5/0.4 mm); (E) Group 1.0 (1.0/0.6 mm) and (F) Group 1.5 (1.5/0.8 mm).

Eschenburg, Germany). Consequently, the negative forms were filled with the resin-based composite in 1.5 mm thick layers and LED-light curing was conducted for 20 s (Bluephase G4 with 1200 mW/cm², Ivoclar Vivadent, Schaan, Liechtenstein). The dentin-like resin dies were then stored in distilled water at 37°C for 3 to 5 weeks in an incubator (Universalschrank UF55, Memmert, Schwabach, Germany) to allow hydration and continuous polymerization.⁵³ Subsequently, all resin dies were embedded in a self-curing epoxy resin (RenCast® CW20/ Ren® HY 49, Huntsman Advanced Materials, TX, USA) to imitate the elasticity of bone.^{54,55}

2.2 | Fabrication of all-ceramic CAD/CAM restorations

The six master preparation dies were scanned in occlusion of the upper and lower typodont models using a laboratory scanner (PrograScan PS5, Ivoclar Vivadent). All reconstructions were designed with their respective thicknesses in a dental CAD software (exocad Dental CAD, exocad, Darmstadt, Germany). The virtual spacer was set at 50 µm. Datasets were calculated in a CAM software (PrograMill CAM Software, Ivoclar Vivadent) and milled out of partially crystallized lithium disilicate CAD/CAM blocks (IPS e.max CAD LT, Ivoclar Vivadent) in a five-axis milling machine (PrograMill PM7, Ivoclar Vivadent AG). Finally, crystallization and glazing of all restorations were performed (Programat EP 3010, Ivoclar Vivadent). All restorations were fabricated by the same master dental technician according strictly to manufacturer's recommendations. Before cementation, the restorations' thicknesses were verified with a caliper (Kroeplin GmbH, Schlüchtern, Germany).

2.3 | Adhesive cementation

Prior to adhesive cementation, the intaglio surfaces of all lithium disilicate restorations were etched with 4.9% hydrofluoric acid (IPS Ceramic Etching Gel, Ivoclar Vivadent) for 20 s and afterwards thoroughly rinsed with air-water spray and dried with oil-free air stream. Subsequently, a primer was applied for 60 s (Monobond Plus, Ivoclar Vivadent).

Resin dies were first pretreated with pumice powder (Picodent, Wipperfurth, Germany), rinsed with air-water spray, dried with an oil-free air stream, and cleaned with 70% ethanol. Afterwards dentin-analogue abutments were pretreated with a light-curing dental adhesive (Adhese Universal, Ivoclar Vivadent). The adhesive was applied for 20 s on the surface of the resin die, gently air-dried and light-cured for 20 s (Bluephase G4 with 1200 mW/cm², Ivoclar Vivadent).

All pretreated restorations were then adhesively bonded with a dual-curing composite cement (Variolink Esthetic DC, Ivoclar Vivadent) to the resin dies. Excess cement was carefully removed with foam pellets and LED light-curing was performed for 20 s from each side (Bluephase G4, Ivoclar Vivadent) under oxygen protection (Liquid Strip, Ivoclar Vivadent). Specimens were then again stored in distilled water for 24 h at 37°C in an incubator (Universalschrank UF55) to allow further polymerization of the adhesive interface before fatigue testing.⁵⁶

2.4 | Fatigue analysis

All specimens were subjected to cyclic mechanical loading with simultaneous thermocycling (5–55°C in distilled water, dwell time 120 s) in

a mouth motion fatigue simulator (CS 4.8 professional line, SD Mechatronik, Feldkirchen-Westerham, Germany) (Figure 4). An occlusal load of 49 N was applied to the disto-palatal cusp of the restoration for 1.2 million chewing cycles at a frequency of 1.6 Hz. To simulate aspects of natural mastication, steatite spheres ($r = 3$ mm, Hoechst CeramTec, Wunsiedel, Germany) were moved 0.5 mm horizontally downwards from the disto-palatal cusp to the central fissure.^{57,58} This in-vitro test-set up is reported to simulate 5-years of clinical behavior under artificial conditions.^{59–61} Specimens were vertically positioned and loaded without angulation. During thermodynamic loading, specimens were visually inspected, aided by a transilluminating crack detection probe (DIA Stick, I.C. Lechner GmbH & Co KG, Stockach, Germany), for cracks, fractures, or debonding twice a day. A survival and success rate after dynamic loading was computed. Specimens, which were unscathed after fatigue were rated as success and samples with evident cracks or debonding, but still in function (in situ) were rated as survival. Samples with catastrophic bulk fractures (fractures where the entire ceramic restoration was fractured and split into two or more pieces) were rated as non-survival.^{15,62,63}

2.5 | Single-load-to-failure testing

After fatigue, all specimens were subjected to a single cycle load-to-failure test (SLF) in a universal testing machine (Zwick Z010/TN2S, Zwick Roell, Ulm, Germany) (Figure 5). Load was applied through a steel ball with a diameter of 6 mm, which was aligned at the same contact point as during thermomechanical fatigue. The vertical cross-head speed was set at 1.5 mm/min. A video camera was placed to observe all specimens during SLF testing and to follow possible failure evolution of samples. Failure was defined as either a visible crack or fracture or a 20% drop in the maximum load ($F_{\max}$) without an obvious event. Specimens were axially loaded until a failure occurred and the maximum load-to-failure force was recorded with the corresponding test software (TextXpert III, Zwick Roell).

2.6 | Failure analysis

Samples were first analyzed using a polarized light microscope (AxioZoom V.16, Zeiss, Oberkochen, Germany) to determine the mode of failure and fracture origin. Z-stack mode (ZEN Core 3.3, Zeiss) was used to gather multiple images with different depths and merge layers within the same image to increase the depth of field.

Most representative fracture types were further subjected to qualitative fractographic analysis using environmental scanning electron microscopy (Quanta 600 FEG ESEM).

2.7 | Statistical analysis

A power calculation (G*Power 3.1.9.7, Düsseldorf, Germany) for a 2×3 factorial design with factors (i) restoration design

(PCR vs. crown) and (ii) ceramic layer thickness (0.5, 1.0, and 1.5 mm) was performed with respect to statistical testing via ANOVA. A sample size of $n = 12$ per group ($n = 72$ in total) was determined to enable detecting effects of at least medium size (Cohen's effect size of $f = 0.26$) with 80% power and a two-sided type-I error of $p < 0.05$ for the two factors and their interactions.

Data were statistically analyzed with SPSS 28 (IBM Corp., Armonk, NY, USA). The Levene Test was applied to test for homogeneity of variance before using ANOVA to test for main effects and interactions of the two factors of interest (restoration design and ceramic layer thickness). The influence of ceramic layer thickness was further analyzed post-hoc using Tukey tests following separate one-way ANOVAs per restoration design. The influence of restoration design was separately analyzed post-hoc for each layer thickness level via two-sample t-tests. The level of significance was set at $p < 0.05$ (95%-CI) for all tests. Data were graphically presented in boxplots.

3 | RESULTS

3.1 | Cyclic fatigue loading

All restorations showed wear scars in the contact area after fatigue exposure, as a result of the movement of the steatite ball during cyclic mechanical loading. The overall survival rate was 100% for all lithium disilicate reconstructions. No debonding of reconstructions occurred during or after dynamic fatigue loading, but not all specimens withstood fatigue application unscathed. Eight crowns of Group C-0.5 (after 35,025, $2 \times 43,757$, 112,196, 386,388, 396,393, 444,262, and 901,591 cycles) and one restoration of Group PCR-0.5 (after 362,856 cycles) showed evident cracks after fatigue (Figure 7).

The overall success rate yielded in 87.5%. The success rate for PCRs was 96.88% and for crowns 75%. The success rates of PCRs and crowns according to ceramic layer thickness are displayed in Table 1.

3.2 | Single load to failure

The results obtained from the SLF testing are summarized in Table 2 and graphically displayed in Figure 6.

Both restoration design (PCR vs. crown) and ceramic layer thickness (0.5 mm vs. 1.0 mm vs. 1.5 mm) showed significant main effects with respect to failure load (preparation design: $F(1.66) = 19.76$, $p < 0.001$, thickness: $F(2.66) = 22.91$, $p < 0.001$). However, a statistically significant interaction between the two factors (restoration design and ceramic layer thickness) could not be detected ($F(2.66) = 1.24$, $p = 0.296$).

Regarding the comparisons of ceramic layer thickness levels separately for each restoration design, post-hoc Tukey tests revealed significant differences for crowns, which were in favor of thicker crowns compared to thinner crowns (C-0.5/C-1.0 $p = 0.005$, C-0.5/C-1.5 $p < 0.001$, C-1.0/C-1.5 $p < 0.001$). However, no significant differences were detected for layer thicknesses of PCRs (PCR-0.5/PCR-1.0

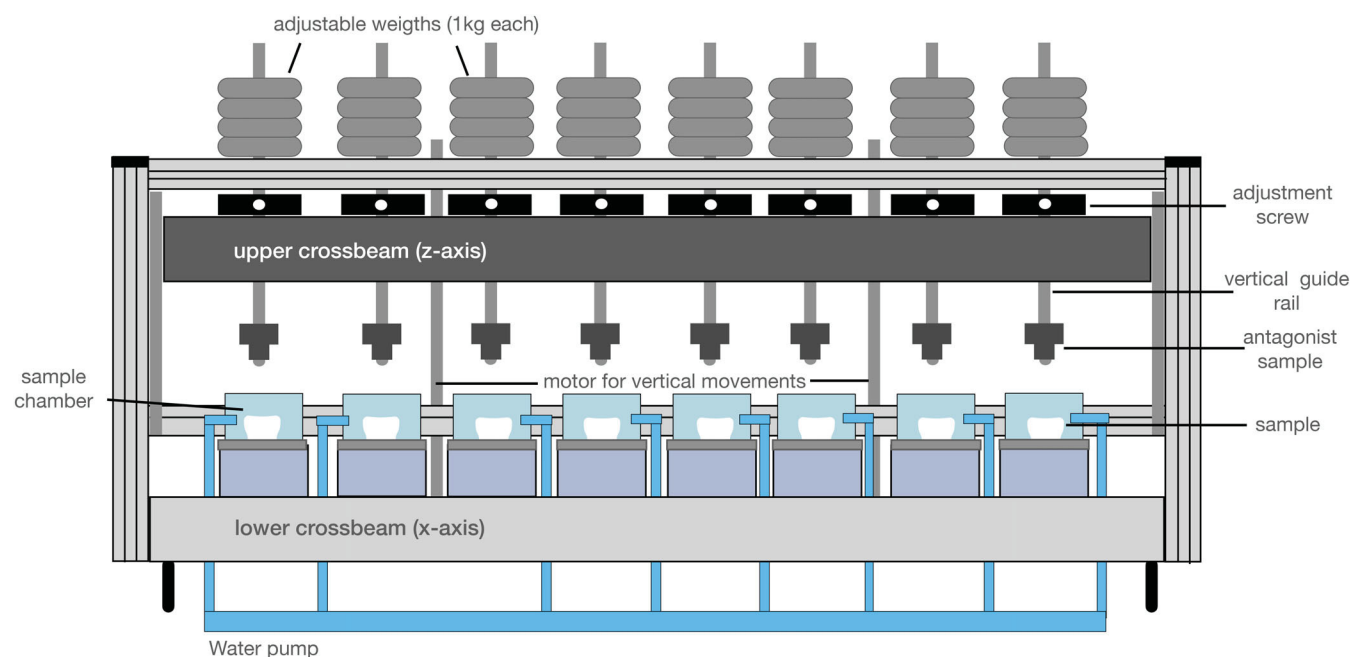


FIGURE 4 Schematic drawing of the fatigue simulator (CS 4.8 professional line, SD Mechatronik, Feldkirchen-Westerham, Germany) with eight sample chambers and simultaneous thermocycling. A total load of 5 kg (=equals 49 N) was applied to the samples, as the vertical guide rail and the sample holder weight together another 1 kg.

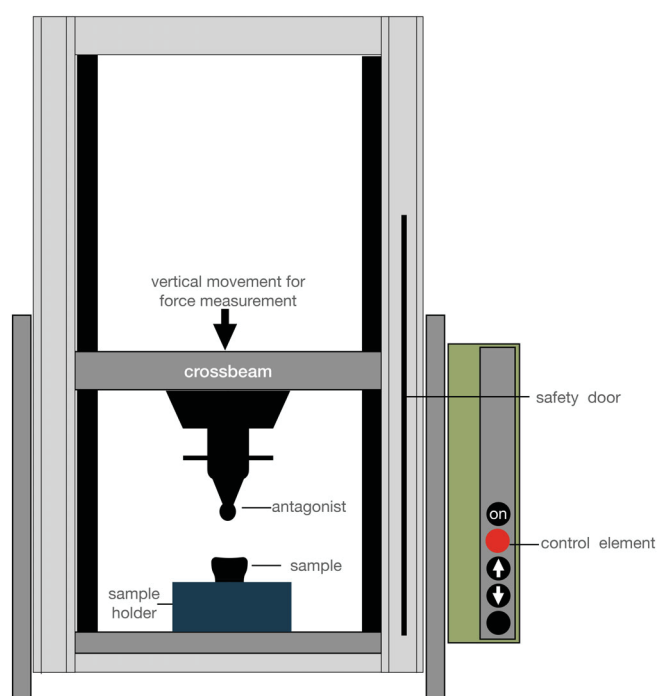


FIGURE 5 Schematic drawing of the universal testing machine (Zwick Z010/TN2S, Zwick Roell, Ulm, Germany) for single cycle load-to-failure testing (SLF).

$p = 0.283$, PCR-1.0/PCR-1.5 $p = 0.277$), aside from the direct comparison for ceramic layer thicknesses of PCR-0.5 and PCR-1.5 where a significant difference in favor of the standard (1.5 mm) thickness ($p = 0.011$) was observed.

Post-hoc *t*-tests for separate comparisons between restoration designs for each level of ceramic layer thickness with respect to restoration design (PCR vs. C) revealed significant differences for thicknesses of 0.5 mm (C-0.5/PCR-0.5: $t(22) = -4.11$, $p < 0.001$) and 1.0 mm (C-1.0/PCR-1.0: $t(15.43) = -3.18$, $p = 0.006$) in favor of PCRs. PCRs with a minimally invasive design and layer thickness outperformed crowns with the identical ceramic thickness. No statistically significant difference between PCRs and crowns was found for a 1.5 mm ceramic layer thickness (C-1.5/PCR-1.5: $t(22) = -1.12$, $p = 0.276$).

3.3 | Failure and fractographic analyses

Failure analysis after fatigue revealed seven samples of Group C-0.5 and one sample from Group PCR-0.5 with radial cracks starting from the intaglio cementation surface beneath the contact point during cyclic loading. The radial cracks spread sideways and upwards. In C-0.5 crowns, radial cracks penetrated through the ceramic layer and progressed relentlessly around the lateral walls under the influence of slow crack growth during moist cyclic loading (Figure 7C,D).^{10,11} However, the radial crack in sample PCR-0.5 remained contained within the ceramic layer (Figure 7A,B). In one sample of Group C-0.5, hoop stresses along the axial walls of the crown lead to crack formation during fatigue.³⁶

After SLF predominately catastrophic ceramic bulk fractures of both PCRs and crowns with involvement of the dentin-analogue dies for thickness of 1.5 and 1.0 mm into 2–3 main fragments were observed. Groups PCR-1.5, PCR-1.0, and C-1.5, showed exclusively

TABLE 1 Success rates after simulated fatigue exposure of 5 years.

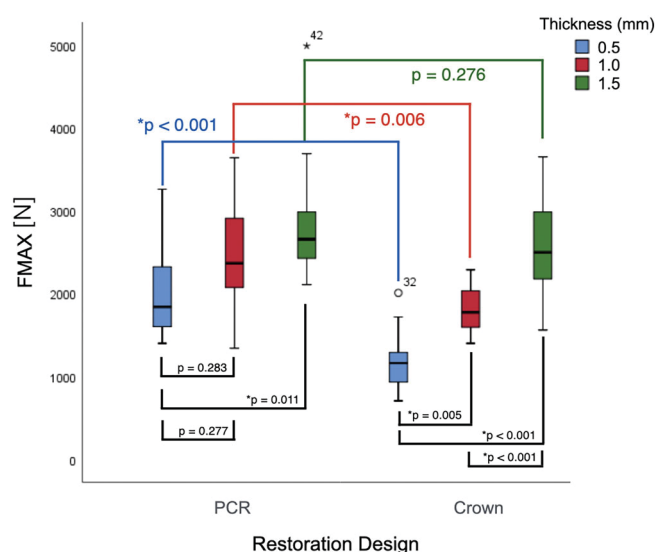
Group	Intact and unharmed specimens after fatigue	Success rate	Success rate of restoration design	Overall success rate
PCR-1.5	12/12	100%	PCRs	87.50%
PCR-1.0	12/12	100%	96.88%	
PCR-0.5	11/12 (one crack after 362,856 cycles)	91.66%	35/36	
C-1.5	12/12	100%	Crowns	
C-1.0	12/12	100%	75.00%	
C-0.5	4/12 (eight cracks after 35,025 cycles, 2 × 43,757 cycles, 112,196 cycles, 386,388 cycles, 396,393 cycles, 444,262 cycles and 901,591 cycles)	33.33%	(28/36)	

Group name	Min	1st Qu	Median	Mean	3rd Qu	Max	SD
PCR-1.5	2210	2386	2660	2883	3380	4990	782
PCR-1.0	1340	2041	2370	2451	2861	3640	646
PCR-0.5	1400	1645	1840	2023	2400	3260	594
C-1.5	1560	2186	2500	2566	2946	3650	598
C-1.0	1400	1610	1775	1798	1986	2290	296
C-0.5	708	959	1165	1193	1428	2010	369

TABLE 2 Failure load results of all tested groups [$N = \text{Newton}$].

Note: 1st Qu = 25% of data was below this value; Median = 50% of data was below this value; 3rd Qu = 75% of data was below this value.

Abbreviations: Max, maximum; Min, minimum; SD, standard deviation.

**FIGURE 6** Boxplot of Failure Loads ($F_{\max}$ in N) of tested Groups (* indicates outlier for Group PCR-1.5 and 0 indicates outlier for Group C-0.5). Statistical significance ($p < 0.05$) is indicated by asterisk in front of p -value.

catastrophic ceramic bulk fractures (100%) with involvement of the resin abutment die (PCR-1.5: 83.3%, PCR-1.0: 75%, C-1.5: 100%), whereas Group C-1.0 showed mainly ceramic bulk fractures (83.3%) and two ceramic fractures with partial detachment of the crown (16.7%) and an intact die. Samples of Group PCR-0.5 and C-0.5

revealed both mixed catastrophic ceramic bulk fractures (PCR-0.5: 50%, C-0.5: 33.3%) and extended “chipping” fractures within the ceramic (PCR-0.5: 50%, C-0.5: 66.6%) after SLF.

Resin dies remained intact more frequently with a thinner thickness of 0.5 mm (PCR-0.5: 75%, C-0.5: 83.3%), compared to thicker thicknesses of 1.0 mm (PCR-1.0: 25%, C-1.0: 25%), and 1.5 mm (PCR-1.5: 16.7%, C-1.5: 0%).

During detailed fractographic analysis, telltale markings were observed, comprising hackle and arrest lines, which may allow to determine the origin of fracture.

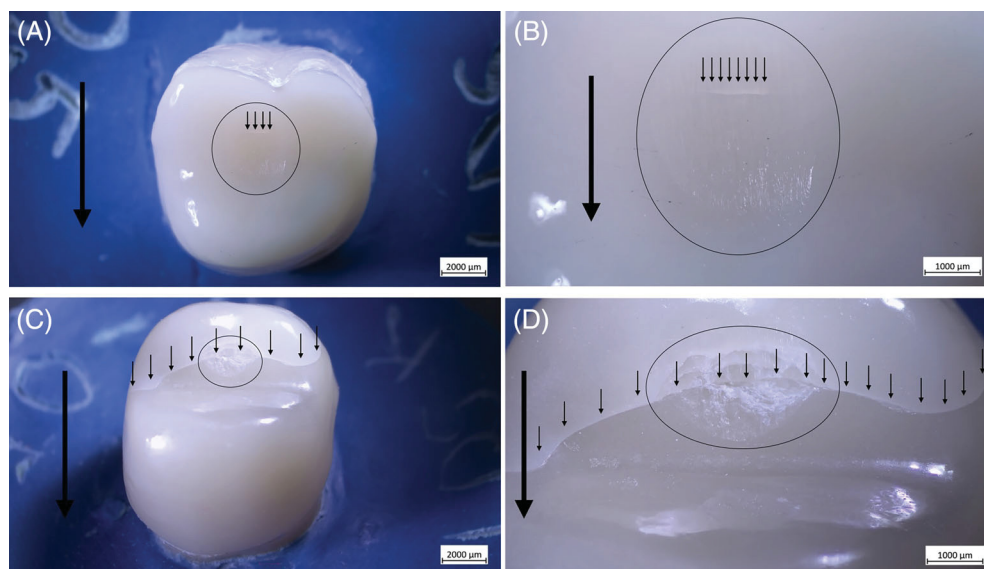
A detailed fractographic analysis of representative specimens of each Group is presented in Figures 8 and 9.

4 | DISCUSSION

The present laboratory study aimed to analyze the influence of restoration design and ceramic layer thickness on fatigue and fracture resistance of posterior monolithic CAD/CAM LDS restorations. The tested research hypotheses were accepted, as both restoration design and ceramic thickness showed significant effects on failure loads.

Clinically, maximum masticatory forces in a range of 289–700 N have been reported for the posterior dentition.¹⁴ Regardless of restoration design and ceramic layer thickness all specimens withstood this occlusal load threshold with mean failure load values of 1193–2883 N. Although not all of the specimens (especially for Group C-0.5) withstood dynamic loading unscathed, all samples were still in

FIGURE 7 CAD/CAM LDS restorations revealing cracks (small arrows) and wear (circle) after fatigue. Bold arrow indicates direction of sliding movement during fatigue. (A + B): PCR-0.5; (C + D) C-0.5.



function and therefore subjected to single load to failure testing as in previous studies.^{24,50,62,63} So far, neither clinical studies nor laboratory investigations could show that pre-damaged reconstructions exhibit a shorter service life or failure loads than non-damaged restorations.^{29,48,63} A possible explanation might be that resulting cracks, depending on their locations, may require much larger driving forces to propagate across an anatomic restoration.⁶ This behavior is similar to cracks observed in human enamel, which do not necessarily result in an immediate increase in failure rates of the compromised teeth.^{37,63}

The success rates of PCRs and crowns were 96.88% and 75%. A systematic review of clinical studies, which compared onlays and full-coverage crowns confirmed high survival rates of 93.5% and 95.38% with no statistical difference between the groups after five years.⁴⁰ Another recently published review with meta-analysis, found equally survival and success rates for both treatment options of onlays/partial crowns and complete crowns after one and 3 years of service.⁴¹ Clinical long-term studies of heat pressed and CAD/CAM-fabricated monolithic LDS crowns, occlusal veneers with a shoulder preparation and partial-coverage restorations reported also high survival rates of 80.1%–96.75% after 15–16.9 years, 100% after 11–13 years and 95.27% after 16.9 years of function.^{29,31,33,48}

The results of the present in-vitro study showed that PCRs revealed higher failure load values compared to complete coverage crowns. Minimally invasive PCRs outperformed the corresponding full-coverage crowns with the identical layer thickness for ultrathin (0.5 mm) and thin (1.0 mm) reconstructions. For standard thick restorations (1.5 mm), the restoration design does not seem to have a significant influence on failure loads. A possible explanation might be due to the complexity of the different restoration designs, as any loss of tooth structure weakens the tooth-restoration system.²¹

Direct comparisons to previously published in-vitro studies in the literature are difficult, since the investigated preparation design of full-veneers on molars has not been analyzed elsewhere. Moreover,

methodological differences in study set-ups and fatigue protocols can highly impact the outcome of obtained fracture strength values.¹⁶

Equivalent failure load values for CAD/CAM fabricated 1.0 mm LDS occlusal veneers with a rounded shoulder or a minimally invasive chamfer of 2395 and 2408 N were recorded after simulated artificial aging (1.25 million cycles, 50 N, 1 Hz).¹⁷ A study which compared standard thickness (1.5 mm) CAD/CAM LDS molar crowns to two overlay designs with a chamfer, observed lower fracture forces of 1018 N for crowns and 436–813 N for the overlays.²³ However, specimens were subjected to a higher load of 275 N and a higher number of loading cycles (2 million, 1 Hz) during aging.²³

Monolithic CAD/CAM fabricated LDS molar crowns were also tested in different ceramic layer thicknesses (0.5–1.5 mm) and various fatigue protocols.^{28,34,39,47}

Lower and partly unsatisfactory failure load values of 470 N (without fatigue) and 369 N (with fatigue) for 0.5 mm, 801 N (without fatigue) and 889 N (with fatigue) for 1.0 mm, 1107 N (without fatigue) and 980 N (with fatigue) for 1.5 mm LDS molar crowns were observed with a similar aging protocol (1.2 million cycles, 108 N).⁴⁷ Yet, the study loaded the specimens in a 30° angle, perpendicular to one cusp and with a higher load during fatigue (108 N) which might explain the lower fracture values.⁴⁷ Two studies where no cyclic fatigue protocol was applied recorded lower failure load values of 827 N (0.5 mm), 1228 N (0.7 mm), 1503 N (1.0 mm) and 1440–1499 N (1.5 mm) for non-aged monolithic CAD/CAM LDS crowns as in the present study.^{34,39} The same resin material (Z100) was used as an abutment in one of the studies simulating the elastic modulus of dentin as in the present investigation.³⁹ A study with a prolonged fatigue application (5 million cycles, 275 N) lead to slightly lower failure load values of 980 N (0.8 mm), 1162 N (1.0 mm) and 1540 N (1.5 mm) compared to the obtained results of the present study.²⁸

Ceramic bulk fractures are still the most common complication for onlays, PCRs, and crowns in clinical observations.^{40,41} Failure modes after SLF revealed more bulk fractures with involvement of the

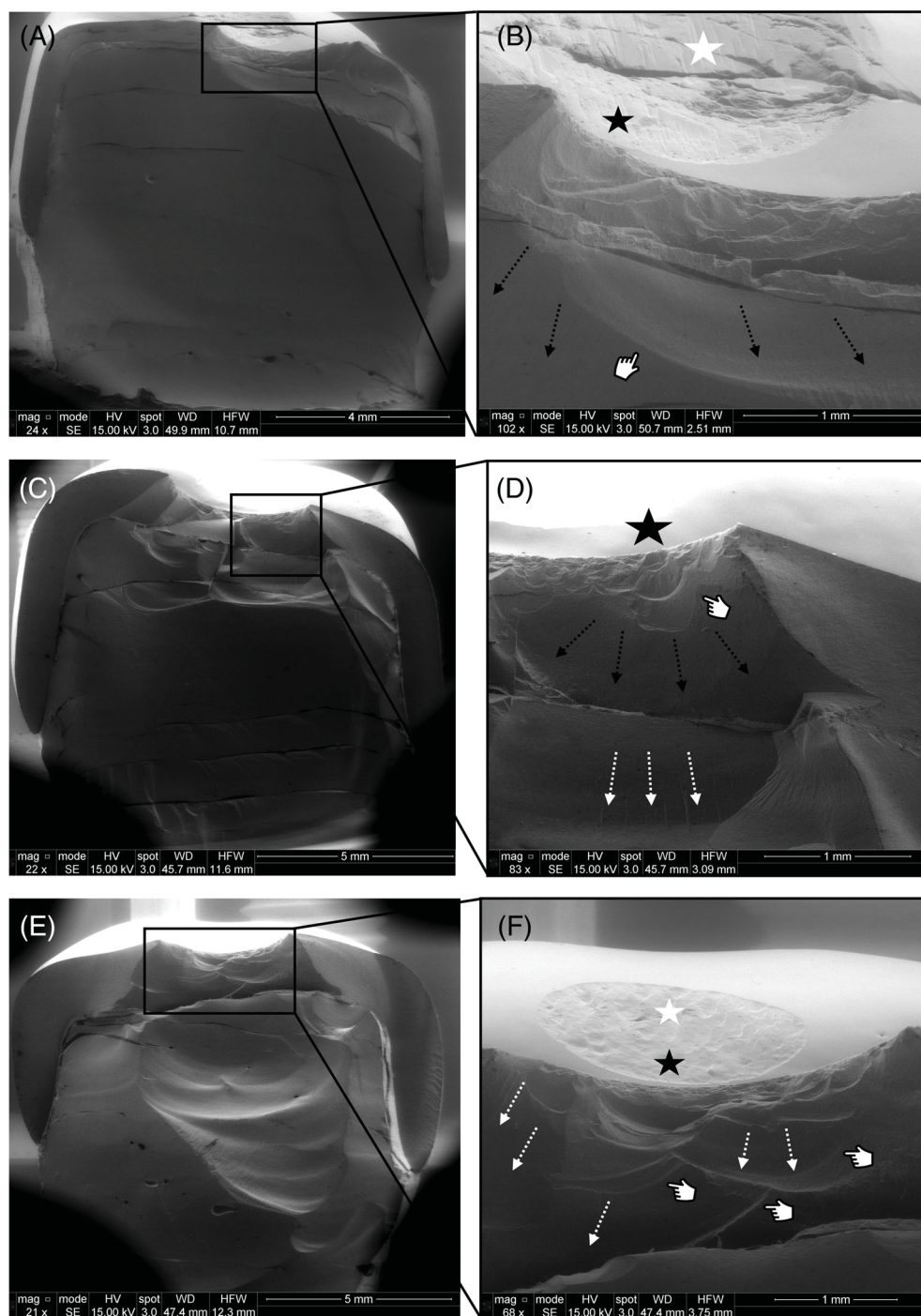


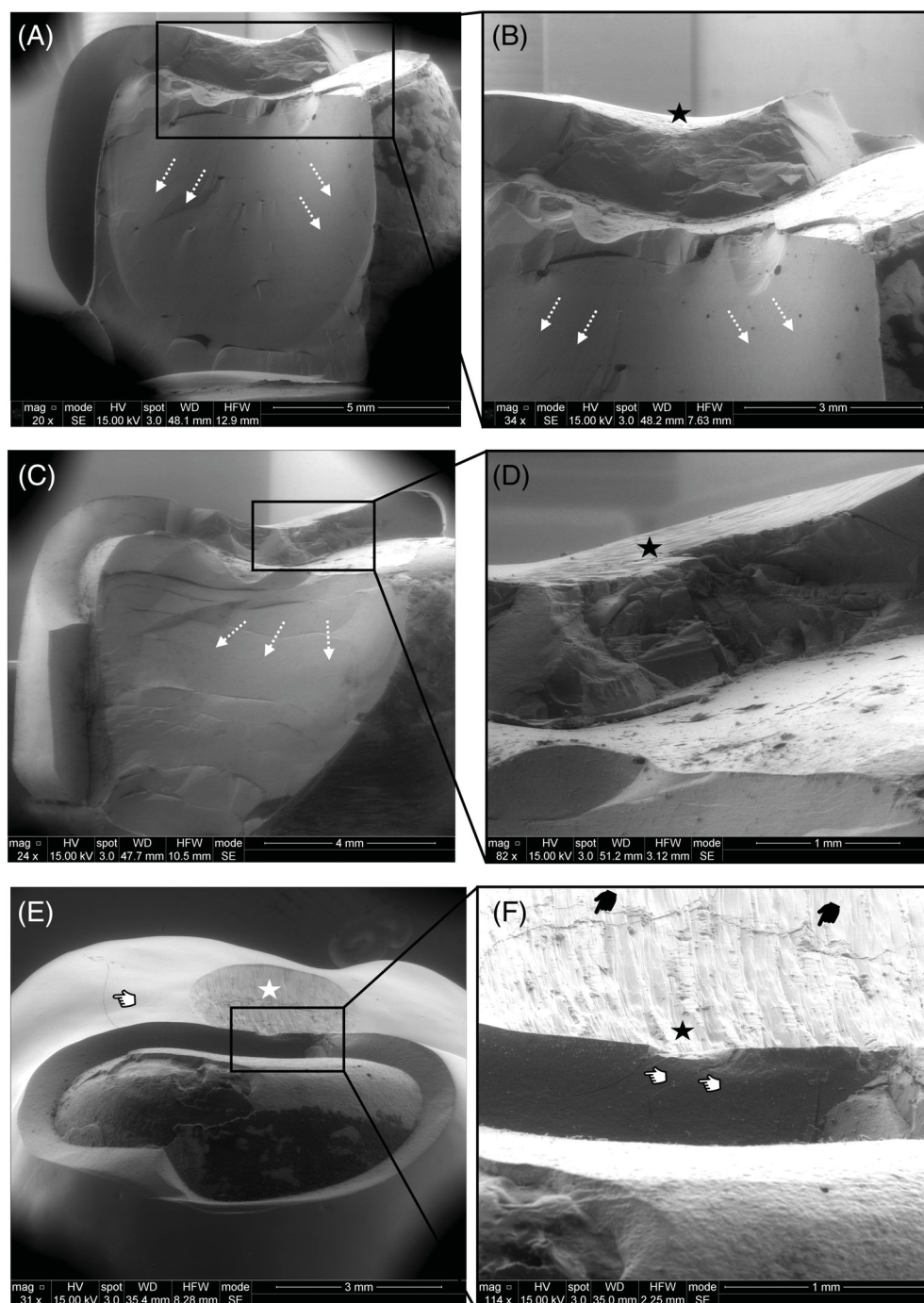
FIGURE 8 Representative scanning electron microscopy (SEM) micrographs of fractured monolithic LDS crowns of C-0.5 (A, B), C-1.0 (C, D) and C-1.5 (E, F). (A, C) and (E) are overviews of the respective LDS crowns and (B, D, F) detailed pictures showing the indentation area (black asterisks) and origin that lead to bulk fracture of the crown with involvement of the underlying resin die. (B) shows sample C-0.5 (same as in Figure 5C,D), with crack (white asterisk) that was already present during fatigue. Dotted lines showing the direction of crack propagation; several arrest lines are visible (pointer) with their concave portion directing toward the origin of failure from the loading area. (D) is a magnification of a specimen from Group C-1.0 where several arrest lines (pointer) and hackle lines (white dotted lines) are visible, which indicate the direction of crack propagation (black dotted lines) from the occlusal surface toward the margins. In (F) surface wear of C-1.5, which occurred during fatigue from the steatite sphere (white asterisk), origin of failure (black asterisk), arrest lines (pointer) and hackle lines (white dotted lines) are clearly apparent.

resin die for thicker layer restorations (both 1.5 and 1.0 mm), whereas with ultrathin (0.5 mm) restorations, failures were restricted to cohesive ceramic fractures without involvement of the dentin-analogue abutment. Accordingly, thinner restorations could provide a “stress shielding protection” for the underlying tooth structure if failure occurs, while thicker restorations could possibly damage the abutment. If the restoration needs to be renewed, this could be beneficial, since tooth substance will be preserved. Other in-vitro studies with minimally invasive LDS restorations on human premolars and molars confirmed the observed failure modes with mainly cohesive ceramic fractures without involvement of the tooth for thin restorations.^{24,49}

In depth fractographic analysis with telltale markings such as hackle lines and compression curls showed that the fracture origin after SLF started for both types of restorations mainly from the loading area. The loading area is the site that had the worst combination of tensile stresses and flaw severity, which then ultimately lead to fracture.¹¹ The evolution of restoration failures occur as a damage competition and accumulation, either resulting from repetitive loading and crack initiation during fatigue or as a result of mechanical overload during SLF.¹¹

Limitations of the present study could be related to the use of composite dies instead of human teeth. This did not allow the

FIGURE 9 Representative SEM pictures of fractured monolithic LDS PCRs of Groups PCR-1.5 (A, B), PCR-1.0 (C, D) and PCR-0.5 (E, F). (A, C) and (E) are overviews of the respective PCRs and (B, D, F) detailed micrographs showing the indentation area (black asterisks) and origin of failure. Specimens of Group PCR-1.5 (A, B) and Group PCR-1.0 (C, D) showed a complete bulk fracture with involvement of the resin abutment, where hackle lines (white dotted lines) indicate the direction of crack propagation. Black asterisk indicates the loading area that lead to fracture of the restoration. (E) and (F) display the same sample of Group PCR-0.5 as in Figure 5A,B with partial detachment of the restoration and “chipping fracture”. White asterisk describes the loading area with radial crack (black pointer) occurred during cyclic fatigue loading. (F) is a magnification of (E) where the white pointers highlight cracks initiating at the occlusal surface and propagating downwards to the cementation surface and resin die, eventually leading to cohesive “chipping” fracture.



investigation and impact of enamel and dentin as a bonding substrate on the load to failure and fatigue behavior of the monolithic LDS PCRs and crowns. However, the utilization of dentin-analogue abutments enabled a high standardization and a systematic analysis of the restorations, without variances in size, age and storage of human teeth. Moreover, a mid-term fatigue protocol was applied instead of long-term fatigue (up to 5 million cycles), which could have potentially led to more cracks or even bulk fractures during cyclic loading. Yet, the employed technique is nevertheless described to be sufficient to detect early failures of reconstructions and widely used in the dental literature.^{3,60}

Since zirconia is presently promoted for minimally invasive tooth-borne restorations, future studies should consider and investigate the proposed restoration design of a full-veneer for this material as well.^{7,12,18} Especially translucent zirconia and multilayer strength-gradient zirconia materials could be interesting candidates for monolithic and minimally invasive posterior reconstructions.

To the best of our knowledge this is the first laboratory study, which systematically investigated the proposed full-veneer/partial-coverage design with monolithic CAD/CAM fabricated LDS restorations on molars with different ceramic layer thicknesses. Full-veneer restorations can be a valuable addendum in the rehabilitation of

posterior dentition, especially for full-mouth cases, caused by advanced tooth wear or bio-corrosive defects. Occlusal and cervical defects that need to be restored can be combined in a minimally invasive single all-ceramic restoration.²⁴

The results of this in-vitro study highlight the superior mechanical performance of minimal invasive full-veneers in comparison to single crowns. Minimally invasive full-veneers should be given preference over full-coverage crowns, whenever possible according to defect configuration. However, it must be noted that manufacturer's recommended minimum layer thickness of adhesively luted LDS crowns is 1.0 mm.

Three-dimensional extended finite element analysis (XFEM) might further enlighten the difference in stress distribution and crack propagation behavior in full-veneers and their beneficial mechanical behavior in comparison to crowns.⁶ Moreover, clinical trials in a prospective or randomized setting are needed to confirm the results of this in-vitro study.

5 | CONCLUSIONS

Based on the results of this in-vitro study, it was concluded that:

The restoration design (partial-coverage restoration vs. crown) and the ceramic layer thickness (1.5, 1.0, and 0.5 mm) affected the failure load of posterior CAD/CAM lithium disilicate restorations.

Minimally invasive full-veneers resulted in higher failure load values than full-coverage restorations.

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CONFLICT OF INTEREST STATEMENT

The authors declare that they do not have any financial interest in the companies whose materials are included in this article.

DATA AVAILABILITY STATEMENT

The data that support the findings of this study are available from the corresponding author upon reasonable request.

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

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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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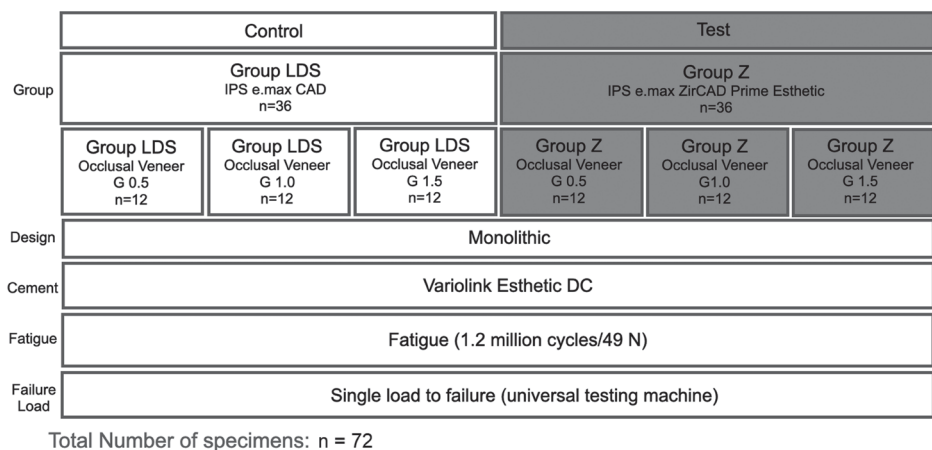


Fig. 1. Test-set up. LDS: lithium disilicate; Z: composition-gradient multilayered zirconia.

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, Tettngang, 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, Wipperfurth, 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/cm², 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 PSS 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 crystallized 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, Wipperfurth, 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/cm², 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-

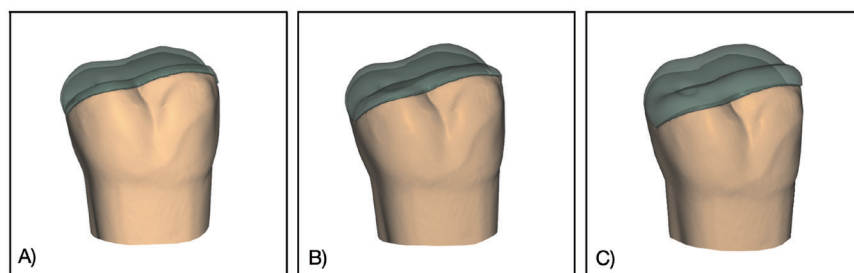


Fig. 3. Restoration design of occlusal veneers in respective thicknesses; A) 0.5 mm, B) 1.0 mm, and C) 1.5 mm.

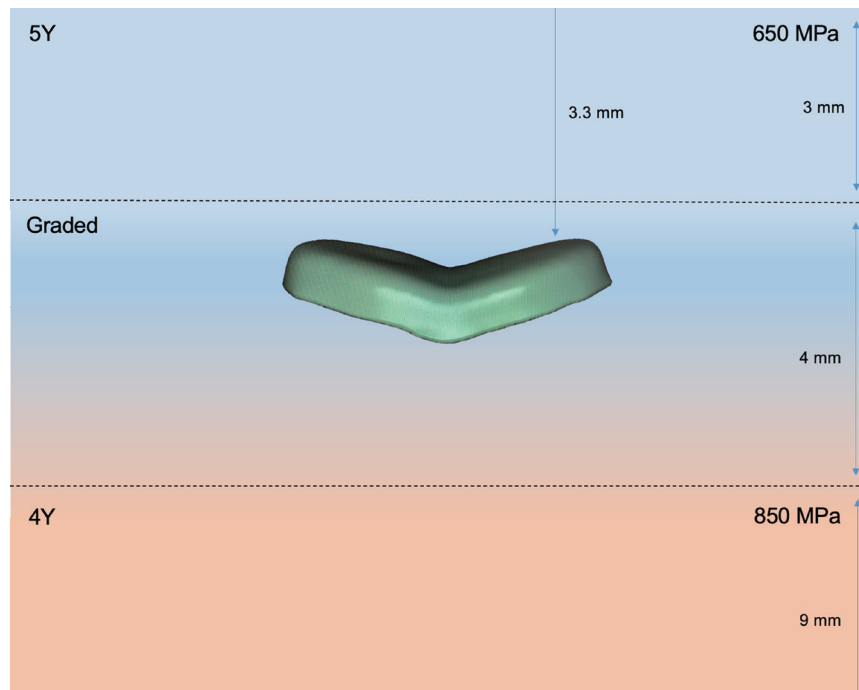


Fig. 4. Veneers of each thickness (0.5 mm, 1.0 mm, and 1.5 mm) were placed within the transition layer of the gradient multilayered zirconia blank, maintaining a distance of 3.3 mm from the upper border of the blank to the top surface of the veneer.

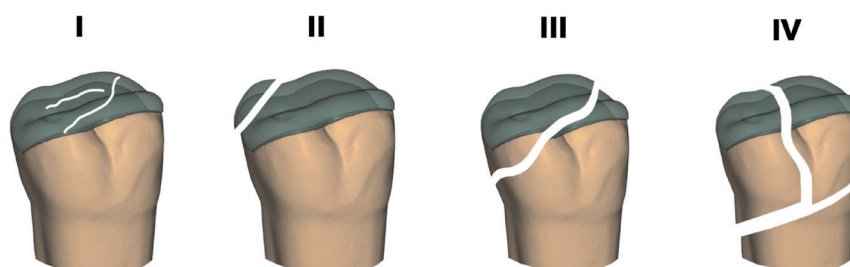


Fig. 5. Failure modes after single load to failure testing: (I) Crack formation within the ceramic. (II) Cohesive fracture within the ceramic, intact tooth. (III) Fracture within ceramic and tooth structure. (IV) Serious/longitudinal tooth fracture involving the root.

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 ($r = 3\text{ mm}$) 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 2×3 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

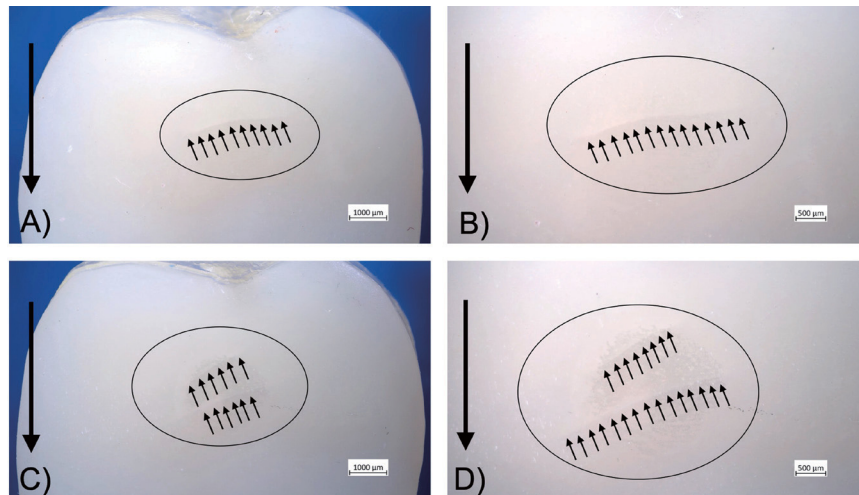


Fig. 6. Light microscope images of 0.5 mm thick gradient multilayered zirconia occlusal veneers (group Z-0.5) showing cracks (small arrows) and wear facets (circle) after fatigue exposure. The bold arrow illustrates the sliding direction during fatigue exposure. A+B): Z-0.5-10 after 381,053 cycles; C+D) 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	1 st Qu	Median	Mean	3 rd 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.

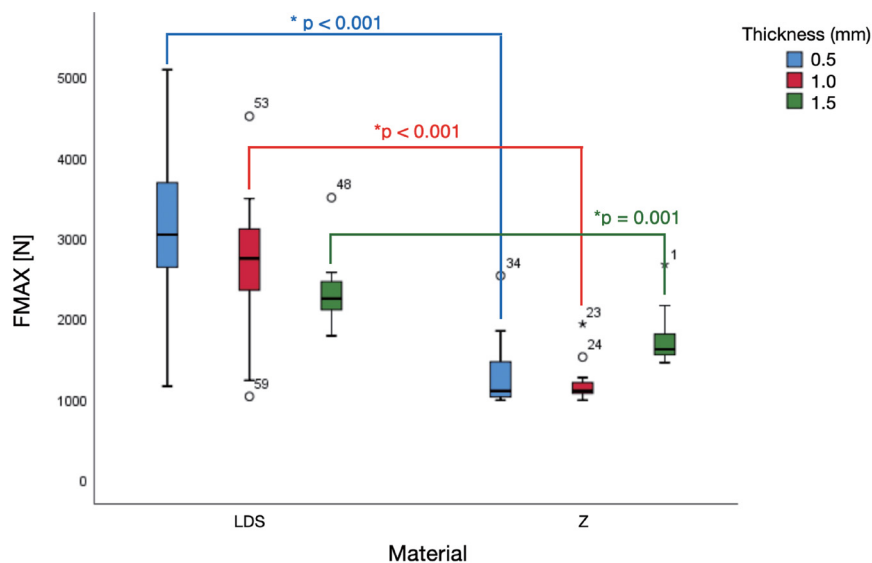


Fig. 7. Boxplot of failure loads ($F_{\max}$ in N) shows statistically significant differences between LDS and composition-graded multilayered zirconia occlusal veneers in different layer thicknesses. Statistically significant results ($P < 0.05$) are marked with an asterisk before the P value. LDS: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

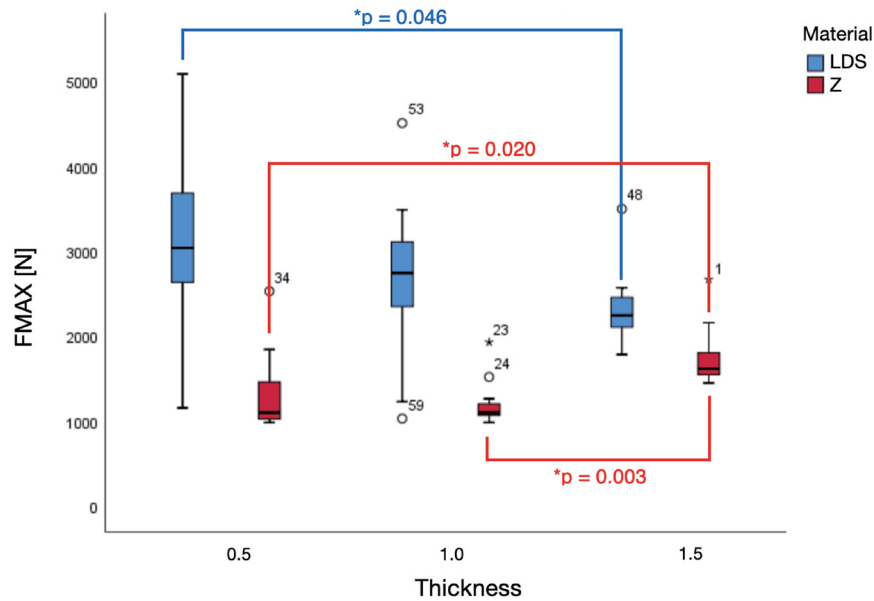


Fig. 8. Boxplot of failure loads (F_{max} in N) shows statistically significant differences between different thicknesses within the test groups (LDS and Z). Statistically significant results ($P < 0.05$) are marked with an asterisk before the P value. LDS: lithium disilicate; Z: composition-gradient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

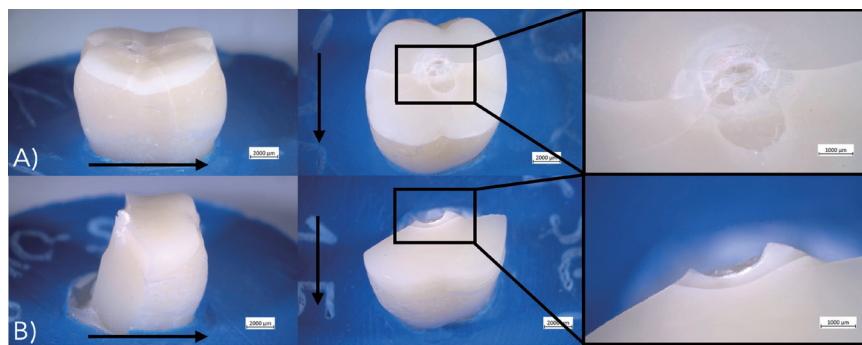


Fig. 9. Light microscope images showing failure modes after single load to failure testing. Ceramic fractures with involvement of the dentin-analogue dies were identified as the dominant failure modes. A) Failure mode III: Fracture within ceramic and tooth structure. B) Failure mode IV: Serious/longitudinal 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 micro-cracks, 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 etch-

ing 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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Article

Failure Prediction of Lithium Disilicate and Composition-Gradient Multilayered Zirconia Occlusal Veneers: A Fractographic and Theoretical Analysis

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Abstract

This in vitro study aimed to evaluate the fatigue behavior of occlusal veneers (OVs) made of lithium disilicate and composition-gradient multilayered zirconia at different thicknesses, incorporating both experimental and theoretical analyses to predict long-term performance. Seventy-two OVs with ceramic layer thicknesses of 0.5 mm, 1.0 mm, and 1.5 mm were fabricated and adhesively bonded to dentin analog composite abutments. All specimens underwent thermomechanical fatigue testing, involving cyclic loading (49 N, 1.6 Hz, 1.2 million cycles) and thermocycling (5–55 °C), simulating five years of clinical function. Fracture patterns were analyzed using light microscopy and scanning electron microscopy. A fatigue lifetime model based on plate-on-foundation theory and slow crack growth was applied to estimate cycles to radial failure. No complete fractures or debonding occurred. However, 50% of 0.5 mm zirconia OVs developed flexural radial cracks from the intaglio surface, while all lithium disilicate and zirconia veneers ≥ 1.0 mm remained intact. Theoretical predictions closely matched the experimental outcomes, indicating that 0.5 mm zirconia performance aligned with the lower-bound fatigue estimates for 5Y-PSZ. Results suggest that lithium disilicate offers superior fatigue resistance at minimal thickness, while thin zirconia is prone to subsurface cracking. A minimum thickness of 0.7 mm is recommended for zirconia-based OVs.

Keywords: occlusal veneer; ceramic thickness; fatigue fracture; fracture mechanics; zirconia; lithium disilicate



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1. Introduction

In recent years, occlusal veneers (OVs) have emerged as a minimally invasive alternative to traditional onlays, characterized by a non-retentive preparation design and reduced layer thickness [1]. Digital workflows, including intraoral scanning, virtual mock-ups and chairside milling enable the fabrication of ultrathin restorations that deliver excellent marginal adaptation and structural accuracy while conserving tooth tissue [2]. Previous in vitro studies have shown that OVs with a thickness of 0.7 mm or less can be successfully used to restore the occlusal vertical dimension in patients with severe tooth wear [3]. Due to its favorable esthetic and mechanical properties (adequate flexural strength (400–600 MPa), an elastic modulus similar to enamel, wear-friendly behavior and reliable bondability), lithium disilicate glass ceramic is widely used in the fabrication of OVs [4–6]. IPS e.max

CAD (Ivoclar Vivadent, Schaan, Liechtenstein) consists of a glassy matrix mainly composed of SiO_2 , Li_2O , P_2O_5 , ZrO_2 , ZnO , K_2O and Al_2O_3 with lithium disilicate ($\text{Li}_2\text{Si}_2\text{O}_5$) as the predominant crystalline phase [7,8]. A recently published systematic review found that CAD/CAM fabricated composite OV's exhibit fatigue and fracture resistance comparable of lithium disilicate veneers, offering a less brittle, dentin-like alternative [9]. However, the only available clinical study reported 3-year survival rates of 100% for OV's fabricated from lithium disilicate (IPS e.max CAD, Ivoclar Vivadent) compared to 84.7% for those made of resin-matrix ceramics (Lava Ultimate, 3M Espe) [10]. These findings emphasize the importance of selecting an appropriate restorative material.

Dental zirconia ceramics have evolved from a largely uniform material with a single composition and homogeneous microstructure into more advanced, often graded structures, characterized by variations in yttria and alumina content, as well as differences in grain type and size [11]. After mechanical and esthetic modifications of the first- and second-generation of zirconia (3Y-TZP (3 mol% yttria stabilized zirconia polycrystals) with or without 0.25 wt.% alumina sintering additives, respectively), highly translucent third-generation zirconia 4Y-PSZ and 5Y-PSZ (4 mol% and 5 mol% yttria partially stabilized zirconia, respectively) gained significant interest for the fabrication of OV's [12]. With increasing yttria content, the proportion of cubic grains increased (~10–20% in 3Y-TZP, ~40% in 4Y-PSZ and ~50% in 5Y-PSZ) [13]. This resulted in enhanced translucency, but at the expense of flexural strength, which decreased from 1000–1400 MPa in 3Y-TZP to 700–1000 MPa in 4Y-PSZ and 400–800 MPa in 5Y-PSZ [14].

In 2015, multilayered fourth-generation zirconia systems were firstly introduced to mimic the shade gradient of natural teeth through pigmentation; hence, they provide a color gradient but a uniform composition with no difference in flexural strength between the enamel and dentin layers [15,16]. More recently, fifth-generation zirconia has been developed by integrating different zirconia compositions within a single blank to combine high flexural strength in the dentin region with increased translucency in the incisal or occlusal areas [14]. Previous studies have shown that different layers of these fifth-generation multilayered zirconia possess distinct compositions, thereby affecting their mechanical and optical properties and, consequently, the fracture resistance of the final restoration [17,18]. Critical flaws introduced during machining or post-processing handling are located in different areas of the blank depending on the fabrication method, while the type of zirconia (3Y, 4Y or 5Y) at the fracture origin also plays a decisive role in fracture resistance [14]. Consequently, true gradient multilayered zirconia materials introduce new challenges to researchers, dental technicians and clinicians. Moreover, predicting clinical outcomes in terms of fracture resistance is particularly complex, as intraoral conditions expose restorations to liquid media, temperature fluctuations that cause repeated thermal expansion and contraction, and cyclic multi-axial loading that generate complex stress fields [19,20]. Manufacturers' recommendations provide important guidance; however, they are largely based on extensive laboratory testing with only limited clinical evidence available at the time of product introduction. Therefore, restoration design and positioning within the blank must adhere to the manufacturer's recommendations, while also being guided by a fundamental understanding of the selected zirconia material [21]. Clinicians should be aware that any modifications beyond these guidelines fall outside the validated indications.

Fractographic analysis, which investigates fracture surfaces to determine crack origin, direction and failure mode, is increasingly used in dentistry and medicine to assess structural integrity, detect material defects and study fatigue behavior [22]. When combined with textural analysis, it can provide additional insights into surface microstructure and wear patterns. These methods can also complement or, in some cases, serve as alternatives to some forms of imaging, including 3D imaging using Cone Beam Computed Tomogra-

phy (CBCT), intraoral scanning and traditional 2D X-rays, especially when 3D imaging is unavailable, impractical, or as an adjunct for surface-level assessment in post-implantation follow-up [23]. The physical mechanisms of fatigue in ceramic restorative materials, especially in restorations cemented to tooth structures under cyclic masticatory loading, have not yet been thoroughly investigated [24]. In particular, evidence on the fatigue behavior of fifth-generation, composition-gradient multilayered OV is scarce and warrants further research [25]. Therefore, this study aimed to evaluate the thermomechanical fatigue behavior and fracture mechanisms of OVs fabricated from gradient multilayered zirconia and lithium disilicate, bonded to and supported by a dentin analog substrate.

2. Materials and Methods

2.1. Specimen Preparation

Seventy-two OV specimens were fabricated from two ceramic materials: IPS e.max CAD LT A2, Ivoclar Vivadent) and composition-gradient multilayered zirconia (Z, IPS e.max ZirCAD Prime Esthetic A2, Ivoclar Vivadent), each in three different thicknesses (0.5 mm, 1.0 mm, 1.5 mm). A maxillary molar typodont model (Frasaco-model, Frasaco, Tettang, Germany) was used to create standardized, non-retentive master dies, which were replicated using a resin composite (Filtek Z100, 3M ESPE, Neuss, Germany) to simulate dentin analog abutments, with an elastic modulus similar to dentin (~18 GPa). This material is a visible-light activated, radiopaque composite with a BIS-GMA/TEGDMA resin matrix. The filler consists of zirconia/silica with an inorganic filler loading of 66 vol% and particle size range of 0.01–3.5 μm [26]. OVs were designed digitally (Exocad Dental CAD, Exocad, Darmstadt, Germany) and milled (PrograMill PM7, Ivoclar Vivadent) from lithium disilicate blocks or zirconia blanks, with zirconia specimens positioned within the transition layer of the multilayered blank (Figure 1). This alignment should reflect clinical CAD/CAM positioning, where the occlusal surface is located in the more translucent 5Y-PSZ for esthetics, while the adjacent 4Y-PSZ layer contributes to higher fracture resistance. IPS e.max CAD blocks were milled (PrograMill PM7 v98.130, Ivoclar Vivadent) in the pre-crystallized (blue) stage, with a milling time of 38 min. The restorations were subsequently subjected to crystallization firing (Programat EP 3010, P161, Ivoclar Vivadent) to achieve their final mechanical properties and esthetics. Similarly, IPS e.max ZirCAD Prime Esthetic was processed in the pre-sintered state, allowing efficient machining (milling time: 11 min), followed by high-temperature sintering (Programat S2, P3, Ivoclar Vivadent) to obtain its final strength and translucency. Surface treatments were applied according to the manufacturer's instructions [7,27]. For adhesive cementation, a dual-curing resin cement composite (Variolink Esthetic DC, Ivoclar Vivadent) was used, consisting of a urethane dimethacrylate-based monomer matrix with ytterbium trifluoride and spheroidal mixed oxide fillers. The filler loading is ~38 vol%, with particle sizes ranging from 0.04–0.2 μm (mean 0.1 μm) [28]. To simulate five years of clinical use [29–31], all specimens underwent thermomechanical fatigue testing, involving cyclic loading (49 N, 1.6 Hz, 1.2 million cycles) and thermocycling (5–55 °C, 120 s dwell time) in a chewing simulator (CS 4.8 professional line, SD Mechatronik, Germany), with periodic inspection for any signs of failure. Additional details of the methodology can be found elsewhere [32].

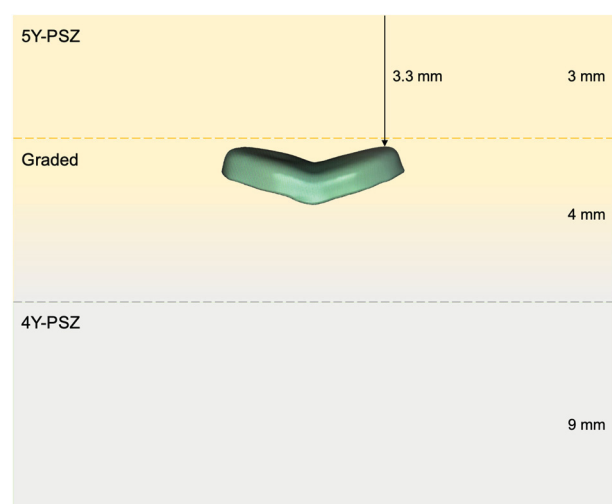


Figure 1. Occlusal veneer (thickness $d = 0.5$ mm, 1 mm, and 1.5 mm) positioned within the transition layer of the gradient multilayer zirconia blank, with a 3.3 mm distance from the upper border of the blank to the top surface of the veneer.

2.2. Fractographic Analysis

For failure analysis, a polarized light microscope (AxioZoom V.16, Zeiss, Oberkochen, Germany) was used to determine fracture origin and failure modes. Additionally, the fracture surfaces of the specimens were examined using an environmental scanning electron microscope (Quanta 600 ESEM, Thermo Fisher Scientific, Waltham, MA, USA). To assess surface involvement, a backscattered electron (BSE) detector was employed. BSE imaging provides material (atomic number, Z) contrast, where heavier elements appear brighter. In this mode, cracks appear as very dark, sharp lines due to the high contrast between the material and vacuum, with the latter appearing nearly black. The smooth, even coloring of the abraded area suggests a uniform material composition and indicates no detectable damage.

2.3. Theoretical Analysis

When a ceramic OV is adhesively bonded to and supported by a compliant dentin abutment analog, two types of fractures may occur due to masticatory contact with the opposing dentition: near-field occlusal surface partial cone cracks induced by frictional sliding contact and far-field cementation surface flexural radial cracks induced by ceramic flexure [33] (Figure 2). The initial occlusal surface partial cone crack sizes are typically in the micrometer range, due to a steep decline in stress concentration beneath the Hertzian contact area [33]. Their propagation is driven by both chemical moist-assisted slow crack growth and mechanical hydraulic pumping fatigue [33]. Therefore, in tough zirconia and glass-ceramic materials, the propagation rate of partial cone cracks is relatively slow, and these cracks are mainly responsible for cusp chip fractures in ceramic restorations [33]. The flexural radial cracks, however, appear as pop-ins on the millimeter scale, thus immediately posing a serious threat of catastrophic failure, especially in thin restorations such as OVs. Here, we focus on the onset of cementation flexural radial fractures in lithium disilicate and zirconia OVs adhesively bonded to a dentin analog substrate under the influence of cyclic fatigue loading, as supported by our current experimental fractographic studies.

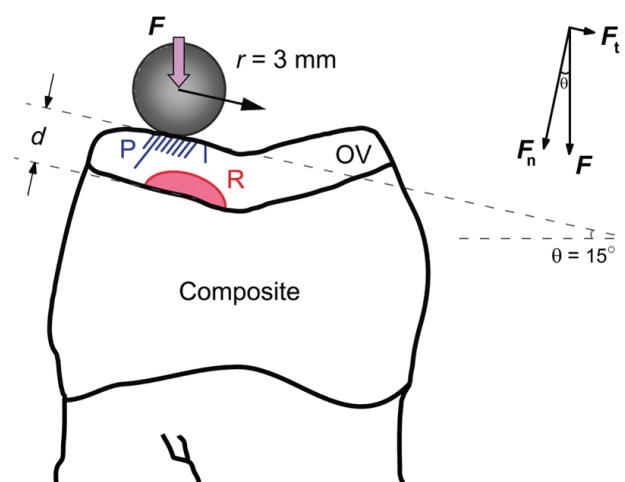


Figure 2. Schematic drawing of load application on occlusal veneer, based on the geometry of a standardized first maxillary molar typodont model. Average thickness d : 0.5 mm, 1 mm, or 1.5 mm; P: Partial cone cracks; R: Radial cracks.

In axial loading on the inner incline of a molar cusp, the applied load F can be decomposed into a normal load F_n and a tangential load F_t (Figure 2). Here, only the normal load component $F_n = F \cos \theta$ contributes to the flexural radial fracture at the cementation surface.

$$F_n = F \cos \theta \quad (1)$$

where $F = 49$ N and $\theta = 15^\circ$ for the current OV designs under the given loading configuration (Figure 2). This yields $F_n = 47.3$ N.

The critical load (F_{nR}) required to initiate a radial fracture from the ceramic intaglio surface was determined using the plate-on-foundation theory, where the ceramic OV acts as a plate supported by a cement/dentin analog foundation. The mathematical model consists of three equations [34,35]:

$$F_{nR} = \frac{B\sigma d^2}{\log\left(\frac{CE}{E^*}\right)} \quad (2)$$

where E is the elastic modulus of the ceramic, zirconia ($=210$ GPa) and lithium disilicate ($=95$ GPa) [36]; d is the thickness of the ceramic; and σ is the flexural strength of the ceramic. For lithium disilicate, $\sigma = 462 \pm 34$ MPa (measured in the authors' lab using a total of 10 samples) [37]. For zirconia, since the OV is fabricated in the graded zone and positioned close to the 5Y-PSZ layer, the substantial 5Y-PSZ content would render its flexural strength similar to that of 5Y-PSZ: $\sigma = 411 \pm 107$ (measured in the authors' lab using a total of 10 samples). C (≈ 1) and B ($=1.35$) are the dimensionless constants [38]; E^* represents the effective modulus of the adhesive cement/dentin analog (Z100) layer, derived from contact mechanics principles [34,39,40]:

$$E^* = E_c \left(\frac{E_s}{E_c} \right)^L \quad (3)$$

where E_s and E_c are the moduli of cement (7.9 GPa, as reported in a scientific document from Ivoclar) and dentin-like substrate (18.6 GPa) [41], respectively. L is a dimensionless function determined experimentally and defined by Equation (4) [34]:

$$L = \exp \left\{ - \left[\alpha + \beta \log \left(\frac{h}{d} \right) \right]^\gamma \right\} \quad (4)$$

where $\alpha = 1.18$, $\beta = 0.33$, $\gamma = 3.13$ [34]; and h denotes the cement thickness, measured in cross-sectioned specimens ($\sim 40 \mu\text{m}$).

However, under fatigue loading, the starting flaw is subject to moisture-assisted slow crack growth prior to instability. This phenomenon is exacerbated by the applied stress intensity and can be described by:

$$\sigma_F / \sigma_0 = (t_0 / t_F)^{1/N} \quad (5)$$

where σ_0 and t_0 are reference parameters related to short-term tests, i.e., the intrinsic flexural strength measured by fast fracture. The time to fracture t_F is given by $t_F = n/f$, where n is the number of cycles to failure and f is the loading frequency (1.6 Hz). N is the velocity exponent of slow crack growth. For a ceramic layer bonded to and supported by a compliant substrate, such values for yttria-stabilized zirconia and lithium disilicate have been previously determined as: $N = 25 \pm 2$ and $N = 20 \pm 3$, respectively [42].

3. Results

3.1. Fatigue Exposure

All restorations exhibited wear scars after fatigue testing, located in the area between the mesio- and distopalatal cusps due to the lateral sliding movements of the steatite ball during cyclic mechanical loading (Figure 3). After fatigue exposure, no debonding or catastrophic ceramic fractures occurred. However, half of the specimens (6/12) of group Z-0.5 (after 95,107 cycles, 95,296 cycles, 381,053 cycles, 630,042 cycles, 656,903 cycles and 840,513 cycles) developed cracks after chewing simulation (Figure 4). All specimens of lithium disilicate and thicker groups of zirconia (Z-1.0, Z-1.5) remained intact (Figure 5). Failure analysis identified cracks of group Z-0.5 as flexural radial cracks starting from the intaglio cementation surface beneath the contact point. However, despite the fact that these radial cracks are over a millimeter long (which is large relative to the thickness of the veneers), they did not propagate to the occlusal contact surface (Figure 6). Parts of these results have been published elsewhere [32].

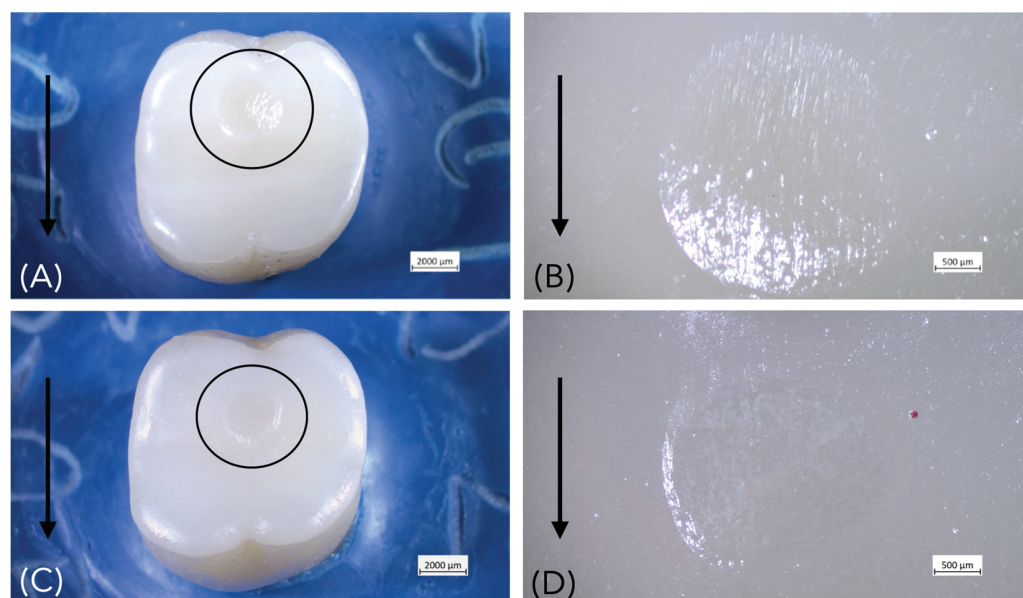


Figure 3. Occlusal veneer restorations following fatigue loading exhibit superficial wear facets (circled) resulting from contact with the antagonist. The bold arrow indicates the direction of sliding during fatigue. (A,B) Lithium disilicate, 0.5 mm thickness; (C,D) Zirconia, 0.5 mm thickness.

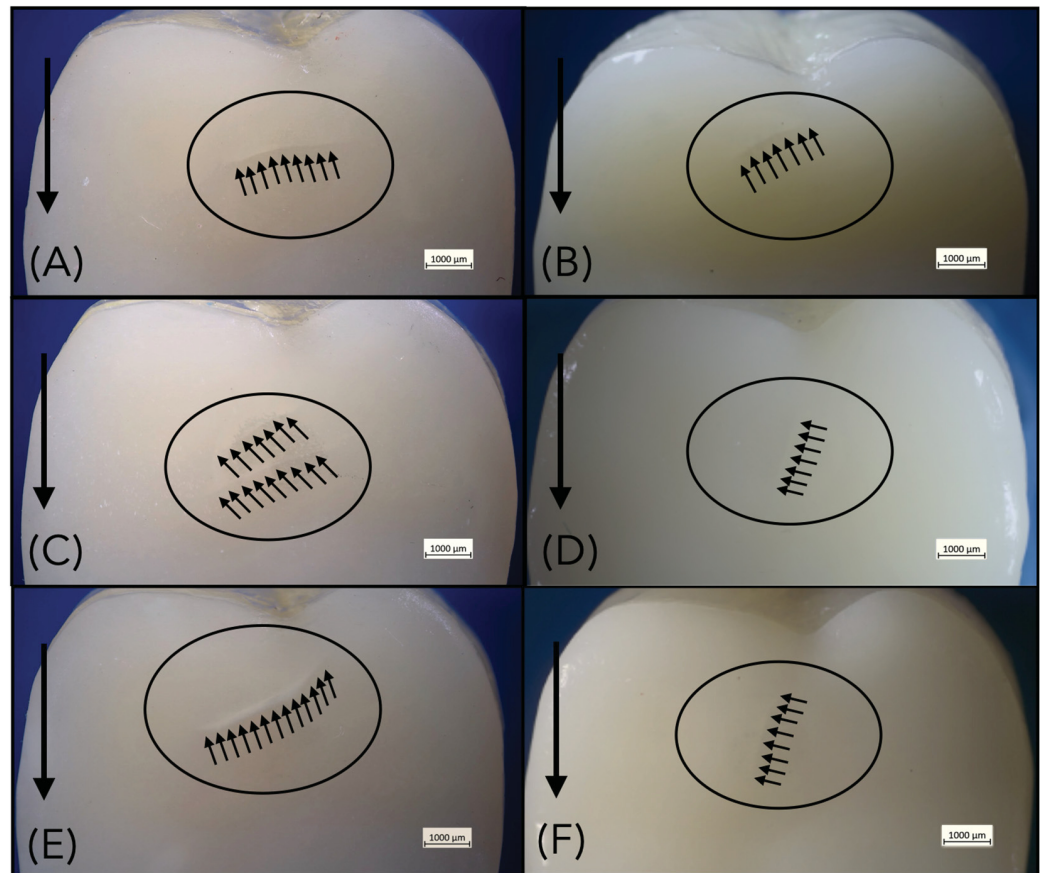


Figure 4. (A–F). 0.5 mm-thick zirconia restorations (group Z-0.5) after chewing simulation. Six out of 12 specimens exhibited cracks (indicated by small arrows) and wear facets (circled). The bold arrow indicates the direction of sliding during fatigue loading.

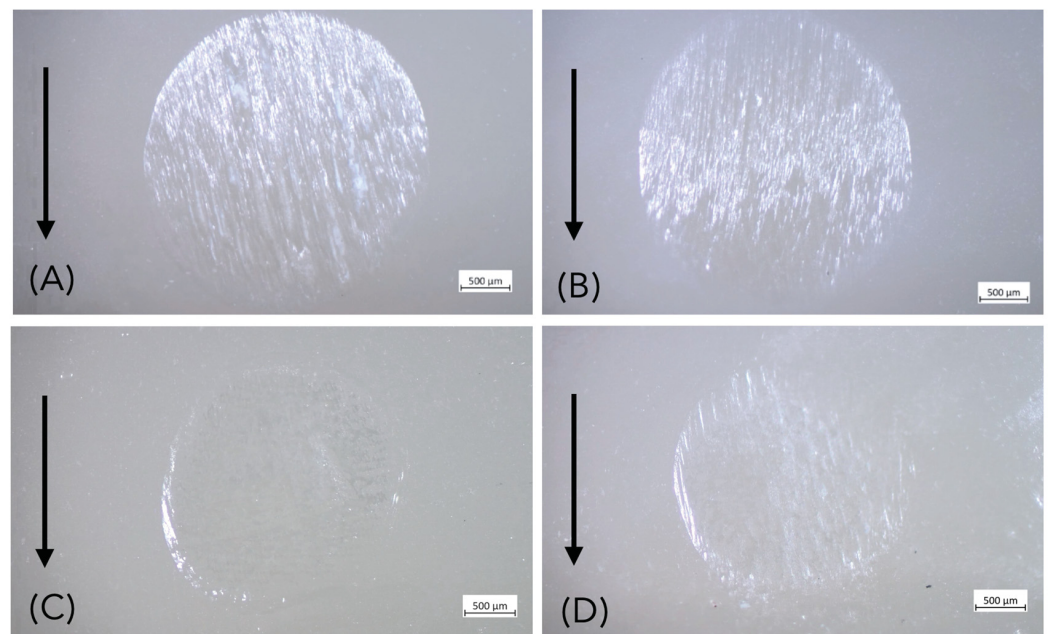


Figure 5. Representative specimens after chewing simulation (A) Lithium disilicate, 1.0 mm thickness, (B) Lithium disilicate, 1.5 mm thickness, (C) Zirconia, 1.0 mm thickness, (D) Zirconia, 1.5 mm thickness. All specimens exhibited wear facets but showed no cracks or fracture features. The bold arrow indicates the direction of sliding during fatigue loading.

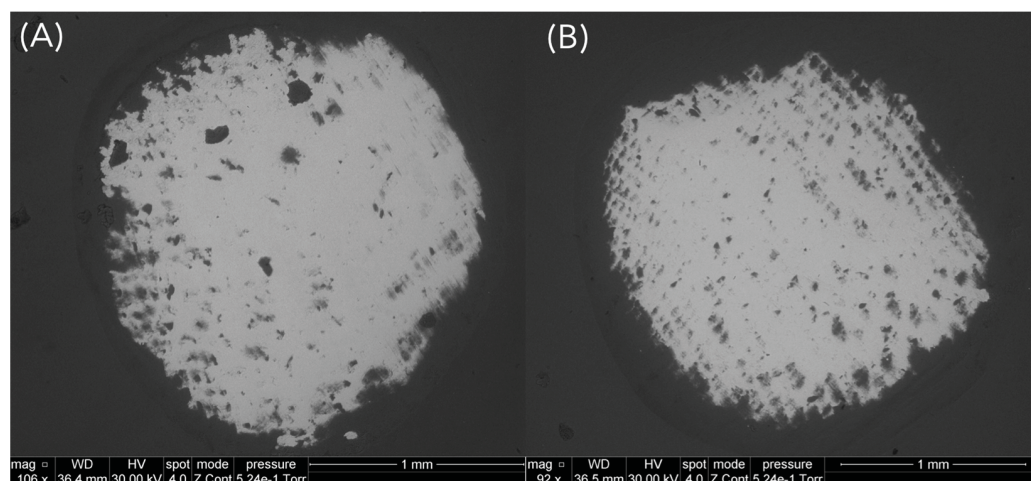


Figure 6. Representative SEM pictures in backscattered electron mode showing samples from group Z-0.5 with wear facets and radial cracks at intaglio surface, indicating the absence of cracks on the outer surface. (A) Zirconia, 0.5 mm thickness, specimen 9; (B) Zirconia, 0.5 mm thickness, specimen 11.

3.2. Calculations

Under cyclic fatigue loading ($F = 49$ N) and utilizing the equations, material properties, and geometrical and dimensionless parameters described in Section 2.3, Figure 7 plots the number of cycles to fracture for the two ceramic OV's of varying thickness. The shaded gray area indicates the predicted lifetime of the zirconia OV's. The dashed line on the left represents the lower bound of the 90% confidence interval, corresponding to a slow crack growth velocity exponent of $N_{\text{lower}} = 23$, whereas the solid line on the right represents the mean lifetime for $N_{\text{mean}} = 25$. Similarly, the red shaded area indicates the predicted lifetime for lithium disilicate OV's, with the lower 90% confidence bound (dashed line) and mean (solid line) corresponding to $N_{\text{lower}} = 17$ and $N_{\text{mean}} = 20$, respectively. Open circles represent the experimentally determined number of cycles to radial fracture for the 0.5 mm zirconia OV's. As shown, the experimental data for the six fractured zirconia OV's are all scattered around the predicted lower bound for this material, providing strong support for the accuracy of our theoretical prediction.

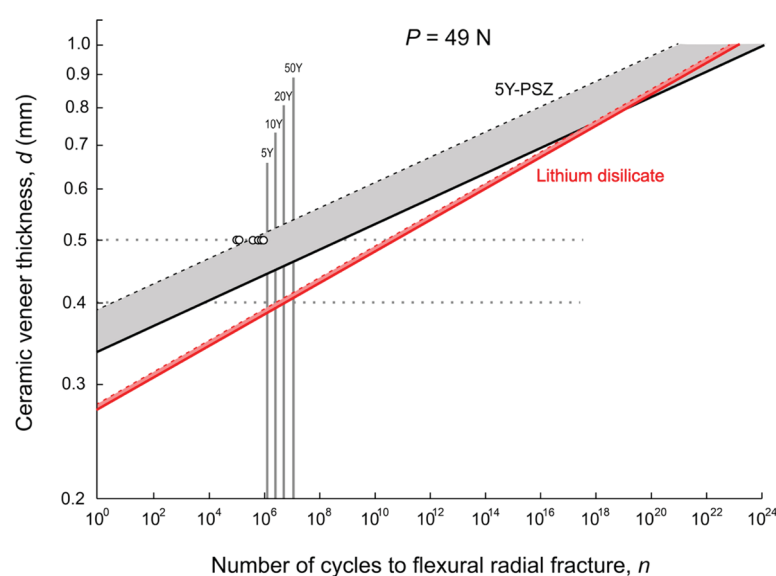


Figure 7. Theoretical prediction of the critical load for radial fracture as a function of the number of cycles for ceramic veneers of varying thicknesses bonded to and supported by a dentin analog composite.

The shaded gray (5Y-PSZ) and red (lithium disilicate) areas represent the predicted lifetime (in number of cycles), bounded by the lower 10% (dashed lines) and mean (solid lines) N values for zirconia ($N_{\text{lower}} = 23$; $N_{\text{mean}} = 25$) and lithium disilicate ($N_{\text{lower}} = 17$; $N_{\text{mean}} = 20$). Experimental data for the six fractured 0.5 mm zirconia OV's (Z-0.5) and their respective number of cycles to fracture are indicated by open circles. Also shown are the estimated years in service, represented by vertical solid lines, based on an assumption of 1.2 million cycles corresponding to approximately five years (5Yrs) [30,31]. Notably, the factor that these six fractured experimental data points are situated around the lower bound of the 5Y-PSZ lifetime prediction provides some validation of our theoretical model.

Based on mechanical properties, 0.5 mm thick lithium disilicate OV's may survive up to 10^{10} cycles—far exceeding any human lifespan. In comparison, OV's made of 5Y-PSZ developed radial cracks even before 10^6 cycles in a 5-year simulation. While 0.4 mm lithium disilicate veneers were estimated to last 10 years, their zirconia counterparts were predicted to fail before reaching 10^4 cycles—well below the 5-year benchmark. However, with increased thickness (>0.7 mm), zirconia veneers may achieve longevity comparable to or even exceeding, that of lithium disilicate.

4. Discussion

Experimental results showed that only the thinnest zirconia group (Z-0.5) developed radial cracks under fatigue, while all lithium disilicate and thicker zirconia specimens remained intact. Computational analysis supported these findings, predicting early failure for 0.5 mm zirconia veneers but long-term durability for lithium disilicate and thicker zirconia restorations. An increasing layer thickness improves fracture resistance through two key mechanisms. First, it reduces flexure and membrane stresses under occlusal loads, following a squared relationship with thickness. Second, it lengthens the crack propagation path, particularly significant for radial cracks at the intaglio surface. This effect is especially relevant for 5Y-PSZ OV's, which are more susceptible to radial cracks at reduced thicknesses and may experience significant durability improvements with increased layer thickness. Compared to zirconia, lithium disilicate is more susceptible to long-term weakening due its higher slow crack growth rate, making it more prone to degradation [43]. Slow crack growth occurs when water molecules or other corrosive agents weaken the strained ionic/covalent bonds at the crack tip, leading to a bond breakage that holds the crack walls together [44]. However, slow crack growth is a gradual process that unfolds over time, often taking years before resulting in fracture [33]. A previous scoping review recommended an occlusal veneer thickness of 0.7 and 1.0 mm for various ceramic materials [3]. The present findings support this recommendation, emphasizing that the ceramic layer thickness depends on material selection and should not be reduced below 0.7 mm when using 5Y-PSZ or the transition layer of multilayered zirconia (between 4Y-PSZ and 5Y-PSZ) for OV fabrication.

To date, no studies have investigated OV's fabricated from the transition layer of gradient multilayered zirconia composed of 4Y- and 5Y-PSZ. In a recently published in vitro study, ultrathin OV's (0.3 mm in the fissures/0.6 mm at the cusps) fabricated from different layers of IPS e.max ZirCAD Prime (3Y, 5Y and transition layer, Ivoclar Vivadent) and IPS e.max ZirCAD MT (4Y-PSZ, Ivoclar Vivadent) were subjected to thermocycling (7500 cycles, 5–55 °C) and chewing simulation (1,200,000 cycles, 98N). OV's made of 3Y-TZP, 4Y-PSZ and the transition layer (between 3Y-TZP and 5Y-PSZ) showed favourable fracture resistance. One veneer made of 5Y-PSZ exhibited a chipping and was assessed as failure. Additionally, 7.8% of 5Y-PSZ veneers were rated as partial failures [12]. In the present study half of the ultrathin zirconia veneer specimens (6 out of 12, 0.5 mm) developed flexural radial cracks (>1 mm in size) after chewing simulation, indicating that positioning veneers within the transition layer between 4Y- and 5Y-PSZ does not enhance fracture resistance. Experimentally determining strength gradients in the transition zone presents a technical challenge,

making the exact mechanical properties of this region difficult to ascertain. Since the veneer restorations were placed near the boundary of the 5Y-layer, the mechanical properties of 5Y-PSZ were used for the calculations in the present study.

A thorough understanding of failure mechanisms is crucial for optimizing the clinical application of gradient multilayered zirconia in conservative restorative dentistry. Compared to monolithic zirconia, multilayered zirconia exhibits more complex failure mechanisms and appears to be more prone to radial cracks. This increased susceptibility may be attributed to differences in material composition and mechanical properties across layers, interlayering interfaces and flaw propagation across different layers [14]. Residual stress can develop within the ceramic material during fabrication, leading to interlayer variations that create potential stress concentration zones [45]. Additionally, layer interfaces may act as sites for crack deflection or delamination, contributing to unpredictable failure patterns. Clinically, radial cracks are more dangerous as they remain beneath the surface and close upon unloaded, making them difficult to detect through visual inspection [46]. Beyond their mechanical implications, subsurface cracks may also serve as potential sites of “contamination” in the oral environment. Fluid penetration, bacterial colonization or the ingress of dietary acids through such cracks could compromise the adhesive interface or accelerate degradation of both ceramic and luting cement. Since the radial cracks observed in this study were confined to the intaglio surface, they do not provide an immediate communication pathway to the oral cavity. However, once initiated, such subsurface cracks may propagate toward the outer surface, weaken the adhesive bond or potentially contribute to secondary caries [47]. While no data currently confirm this contamination pathway in occlusal veneers, previous studies have shown that cracks in natural teeth can harbor extensive bacterial colonization [48,49], indicating a clinically relevant risk that warrants further investigation. Maintaining an adequate thickness (>0.7 mm) seems to be essential for the longevity of gradient multilayered zirconia OV restorations, especially given their high elastic modulus but relatively low flexural strength. This requirement is even more critical than in 3Y-TZP and 4Y-PSZ zirconia (higher strength) and lithium disilicate (lower modulus) [3,12]. To minimize the risk of radial crack propagation, it is important to preserve this minimum thickness during preparation and to avoid unintentional reductions during final adjustments of the OV.

Discrepancies in elastic modulus between ceramics and their supporting structures can significantly affect the material's load-bearing capacity [36]. A previous study showed that the load-bearing capacity of lithium disilicate (872 N) is superior to that of 5Y-PSZ (715 N) and even comparable to 4Y-PSZ (864 N), when supported by a dentin-like resin composite. While ultra-translucent 5Y-PSZ zirconia has significantly lower fracture resistance compared to 4Y-PSZ, both zirconia materials possess a similar elastic modulus (~ 210 GPa), which is considerably higher than that of lithium disilicate (~ 95 GPa) and dental hard tissues (enamel: ~ 70 GPa, dentin: ~ 18 GPa) [36]. Comparable analyses have been applied to other dental materials exposed to repetitive occlusal loading, such as 3D-printed resins used for splints and orthodontic devices. In these materials, fractal dimension and texture analysis revealed that rigid resins maintain surface integrity under compression, whereas flexible resins exhibited pronounced structural degradation and crack formation. Although their elastic moduli (~ 0.7 – 2.7 GPa) and fracture patterns differ markedly from ceramics, both materials classes exhibit occlusal force-induced microstructural changes, highlighting the value of fractal and textural analyses in fatigue assessment [50].

Several limitations of this study arise from its *in vitro* design. While *in vitro* testing is essential for assessing biomechanics, it cannot fully replicate the clinical complexity, including variable loading directions, chemical challenges and biofilm activity. To maintain consistency, typodont teeth and resin composite were used as substrates, which minimized

biological variability (e.g., age, morphology, pre-existing defects and previous clinical history) and allowed standardized specimen preparation. Nonetheless, these substitutes do not entirely replicate the mechanical and structural properties of natural teeth. Furthermore, the theoretical analysis is restricted by the material constants embedded in the equations (e.g., elastic modulus, strength parameters), which are assumed to be homogeneous and isotropic. It was also assumed that the ceramic/resin cement composite interfaces are sufficiently strong, so that at the point of ceramic fracture, the entire system remains bonded together. Such implicit assumptions may not fully reflect the variability of the clinical conditions. Since previous investigations [12] on occlusal veneers reported fatigue cracks occurring exclusively in the 5Y-PSZ region, the veneers were positioned in the 4Y/5Y transition zone to explore whether comparable esthetics could be achieved while potentially enhancing fracture resistance. The mechanical properties, i.e., elastic modulus and flexural strength, of this zone were inferred based on its proximity to the 5Y layer, as direct strength measurement across the gradient zone remains technically challenging. This is because the elastic modulus of 5Y- and 4Y-PSZ is essentially the same, while the strength of a composition predominantly consisting of 5Y-PSZ is expected to approximate that of 5Y-PSZ, since fracture typically initiates in the weaker of the two materials. Nevertheless, this assumption introduces some uncertainty but was considered a conservative approach for evaluating worst-case performance. Fractographic analysis in this study was qualitative, aimed at verifying fracture origins and supporting the theoretical modeling approach. A statistical analysis, including comparisons across thicknesses and ceramic materials, was conducted in a preceding experimental study [32], which provided the basis for the present lifetime predictions. Finally, the study was limited to materials from a single manufacturer. While this ensured standardized processing and provided detailed data on material composition and properties required for the theoretical modeling, it may limit the generalizability of the findings. Future studies should therefore include materials from different manufacturers to enable for broader validation.

In clinical situations where high strength is essential and esthetic demands are moderate, such as in the molar region, monolithic zirconia (3Y-TZP or 4Y-PSZ) or lithium disilicate may be preferred over gradient multilayered zirconia. In esthetically demanding regions, OV's fabricated from multilayered zirconia can present a promising and viable restorative alternative. However, with reduced thickness, translucency differences between 3Y-TZP and 5Y-PSZ become progressively less pronounced. To ensure optimal performance and long-term durability, it is crucial to use multilayered zirconia with adequate layer thickness (>0.7 mm).

5. Conclusions

Gradient multilayered zirconia OV's placed in the transition zone between 4Y- and 5Y-PSZ exhibit a higher risk of fatigue-induced radial cracking at reduced thicknesses. In contrast, lithium disilicate veneers demonstrate superior fracture resistance under simulated oral conditions and may offer enhanced long-term performance. To optimize clinical outcomes, zirconia-based translucent OV's should maintain a minimum ceramic thickness of 0.7 mm.

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Abbreviations

The following abbreviations are used in this manuscript:

OV	Occlusal veneer
3Y-TZP	3 mol% Yttria-stabilized Tetragonal Zirconia Polycrystals
4Y-PSZ	4 mol% Yttria Partially Stabilized Zirconia
5Y-PSZ	5 mol% Yttria Partially Stabilized Zirconia
CAD	Computer-Aided Design
SEM	Scanning Electron Microscopy
BSE	Backscattered Electron
DC	Dual Curing
PM	PrograMill
Hz	Hertz
GPa	Gigapascal
MPa	Megapascal
N	Newton
µm	Micrometer
°C	Degrees Celsius

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