3D-Printed Zirconia and Lithium Disilicate in Dentistry and Their Clinical Applications

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This review focuses on the progressive role of 3D printing in dentistry, particularly emphasizing the use of zirconia-based and lithium disilicate (LS_2)–based ceramic materials. Celebrated for their biocompatibility and esthetic resemblance to natural teeth, these materials have shown promising results with high success rates. Digital light processing (DLP) and stereolithography (SLA) have been noted for producing superior 3D-printed ceramic products. Despite facing challenges such as surface defects, mechanical strength limitations, and esthetic inconsistencies, active research is dedicated to refining the quality and esthetics of 3D-printed zirconia-based and LS_2-based ceramics. This review acknowledges the need to mitigate the steep costs of this manufacturing form and recognizes the current shortfall in clinician and technician awareness of these advanced techniques. Addressing these issues through focused research on improving surface quality, dimensional accuracy, and mechanical properties of 3D-printed dental prostheses is crucial, as is enhancing the dental community's understanding and acceptance of this innovative technology. *Int J Prosthodont 2025;38:12–26. doi: 10.11607/ijp.8831*

Additive manufacturing, or 3D printing, has surged in popularity, enabling the creation of 3D structures from computer-aided designs (CAD).[1] In 1986, Charles Hull pioneered stereolithography (SLA), using ultraviolet (UV) light to solidify resin, and advanced 3D-printing systems.[2] Today, SLA allows for creating accurate dental prostheses using materials such as polymers and ceramics.[1,3] Ceramic restorations manufactured using conventional techniques have difficulty achieving the required surface quality and dimensional precision, can be time-consuming, and are associated with high costs.[4,5] These challenges have led to exploring alternative approaches to manufacture ceramics.

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In the 1900s, Marcus and Sachs introduced 3D-printed ceramics as a potential approach.[6,7] Recently, there has been a growing interest in using 3D-printing to fabricate ceramic materials, especially in dentistry. This surge in interest could be attributed to factors such as reduced likelihood of crack propagation, material wastage during the restoration fabrication procedure, and the technique's ability to fecilitate stratification of different materials within the same restoration.[8] According to the American Society for Testing and Materials (ASTM) International Committee F42 on 3D-Printing Technologies, 3D-printing processes are classified into seven categories.[9] However, not all 3D-printing technologies suit 3D printing of ceramic materials.[10] 3D-printing technologies for ceramics can be further categorized into powder, slurry, and bulk-solid processes.[4] Powder-based 3D-printing technologies use ceramic powders. In these processes, ceramic particles are bonded through melting, laser-induced sintering, or binder application. The methodologies include 3D printing, selective laser sintering (SLS), and selective laser melting (SLM).[5,10] Meanwhile, slurry-based 3D-printing ceramic technologies use a liquid or semi-liquid system, often called inks, dispersed with ceramic particles. During this process, ceramics can be 3D printed using vat photopolymerization (SLA or digital light processing [DLP]), material jetting (MJ), or material extrusion (ME) based on the technique used. The bulk solid-based 3D-printing procedure uses filaments or precursor sheets of ceramics as feedstocks.[5] This technology includes techniques such as fused deposition modeling (FDM) and laminated object manufacturing (LOM) for 3D-printing ceramics.

Zirconia and lithium disilicate (LS_2) are emerging as prominent all-ceramic systems. Zirconia is a resilient polycrystalline ceramic possessing notable mechanical and optical properties, such as fracture toughness of 5 to 19 MPa√m and flexural strength ranging from 500 to 1,200 MPa.[11–13] Notably, zirconia resists acid etching, promotes biocompatibility, reduces plaque accumulation, and wears minimally on opposing dental structures. It remains insoluble in water and resists corrosion in oral settings.[13,14] Due to their superior mechanical properties, 3D-printed zirconia objects are sought-after in the aerospace, automotive, medical, energy, and dental sectors.[15,16] LS_2, a glass ceramic, comprises 65 vol% lithium disilicate, small needle-shaped crystals dispersed in a glass matrix, and a 1 vol% porosity. It has reported superior mechanical properties such as flexural strength of 350 MPa, fracture toughness of 3.3 MPa√m, and heat extrusion temperature of 920 °C.[13]

While 3D printing of ceramics for clinical applications is not yet mainstream and most all-ceramic restorations are still milled, understanding the methodologies for novel 3D-printed ceramic substrates, such as zirconia and LS_2, is essential. This review investigated the current advancements in 3D-printing technologies for fabricating zirconia-based or LS_2 materials specifically used for dental applications.

MATERIALS AND METHODS

The authors developed the following research question: "What are the recent advancements in the 3D-printed zirconia and LS_2 materials within the field of dentistry?"

The search for pertinent articles was conducted across three online databases: PubMed, Scopus, and Web of Science. The final search was conducted on December 15th, 2023, without any year-of-publication restrictions. Keywords employed during the search included: (Additive Manufacturing OR 3D-printing) AND (robocasting OR direct ink writing OR material extrusion OR stereolithography OR digital light processing OR material jetting OR direct inkjet printing OR binder jetting) AND (dental materials OR dental applications OR dentistry) AND (zirconia OR zirconia composites OR ceramic composites) AND (leucite OR lithium disilicate OR vitroceramics). The search process uses Mendeley software (version 1.19.8). Two independent researchers (W.S.L. and A.A.) began by assessing the relevance and alignment of titles and abstracts obtained from the initial search against the eligibility criteria. The articles were classified as include, exclude, or uncertain. The researchers then conducted an independent review of the full-text articles categorized as include and uncertain to determine further eligibility. Any disagreements in the screening process, whether at the title/abstract stage or full-text review, were resolved through discussion until consensus was achieved. Following the full-text assessment, articles that did not meet the criteria were excluded, and the rationale for each exclusion was meticulously recorded.

The inclusion criteria for the present study were as follows: in vitro and in vivo studies involving zirconia-based or LS_2-based materials; research employing 3D-printing technologies; inquiries into the properties of the printed materials; studies focused on materials designed for dental applications; and articles published in English in peer-reviewed journals. Clinical reports were included for the 3D-printed LS_2-based materials because the in vitro and in vivo studies were limited. Exclusions were made for literature reviews, systematic reviews, manufacturer reports, and conference abstracts.

RESULTS

From an initial pool of 1,598 articles, 913 duplicates were efficiently removed. A subsequent review of abstracts excluded 614 articles that did not meet the review's primary focus. Upon meticulous examination of the full texts, 21 articles were excluded due to their lack of specificity regarding zirconia, and none addressed LS_2 directly.

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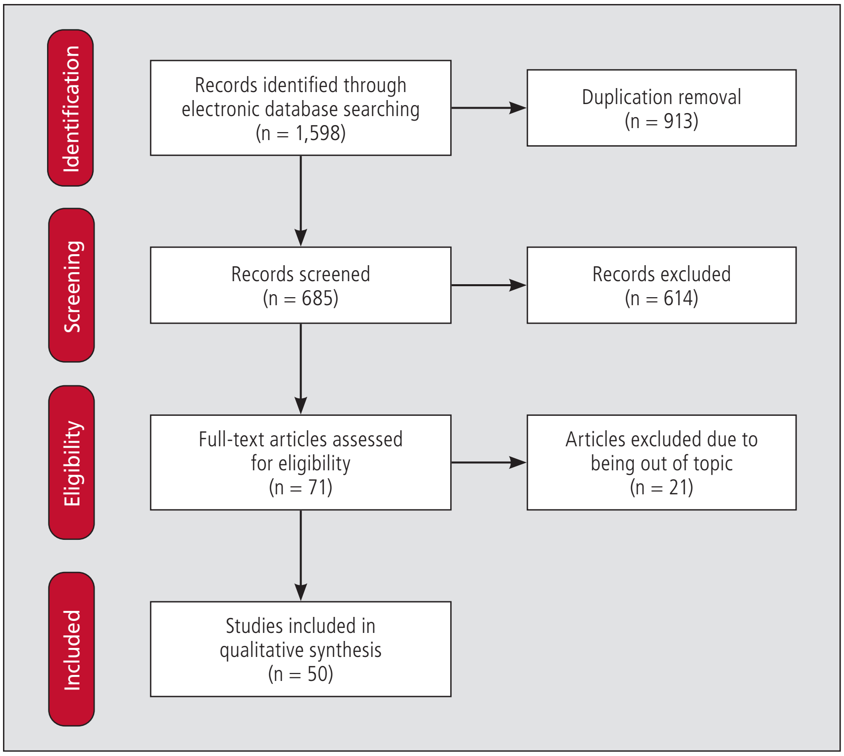

Fig 1 Search strategy (PRISMA flowchart).

Ultimately, this rigorous screening process resulted in selecting 50 articles that satisfied the strict inclusion criteria. Of these, 44 studies focused on exploring 3D printing with zirconia, and 6 articles were dedicated to investigating the 3D printing of LS_2. All 44 studies that examined the 3D printing of zirconia were conducted in vitro. In contrast, out of the 6 studies on the 3D printing of LS_2, 2 were clinical studies, while the remaining 4 were conducted in vitro. The PRISMA flowchart illustrates the detailed selection process (Fig 1). The timeline analysis of publications reveals a growing interest in the 3D printing of zirconia materials, with a distribution as follows: 1 article each in 2011, 2012, and 2014; 2 each in 2015 and 2017; 1 in 2018; 7 in 2019; 10 in 2020; 9 in 2021; and 10 articles in 2022 (Table 1). In contrast, the literature on 3D-printing of LS_2 presents a different pattern, with a single article in 2020, 1 in 2022, and 4 in 2023, highlighting a nascent and rising focus on this particular material in the field of 3D printing (Table 2).

The scope of studies on the mechanical properties of 3D-printed zirconia is broad, encompassing a variety of manufacturing techniques, such as DLP, SLA, MJ, and robocasting (RC). 3D-printed zirconia has been applied to various dental applications, including copings, implants, crowns, and fixed dental prostheses. The evaluated properties include shrinkage, density, microstructure, hardness, flexural strength, surface roughness, trueness, precision, cure depth, fracture toughness, elastic modulus, marginal adaptation, bond strength, dimensional accuracy, deformation under compression, phase composition, antagonist wear, surface deviation, and fracture force pre- and post-fatigue testing. These studies provide insights into mechanical performance trends across various dental applications and manufacturing methods.

The investigation into LS_2 for 3D-printing spanned diverse manufacturing methodologies, including lithography-based ceramic manufacturing (LCM), RC, DLP, and SLA. These studies extensively scrutinized properties such as marginal adaptation, mechanical properties, porosity, shrinkage rate, Vickers hardness, translucency, total layer thickness, flexure strength, Weibull modulus, and elastic modulus. The dental applications examined primarily encompassed veneers and unspecified dental prostheses.

DISCUSSION

3D-Printed Zirconia

Mechanical properties

Several studies evaluated the mechanical properties of 3D-printed zirconia using DLP and SLA technology.[19–21,23,24–28,30–34,39–41,43–46,48–50,52,53,56,58,59] Most of these studies used ceramic powders at various volumetric percentages, ranging from 34.5% to 58.0%. The theoretic densities ranging from 98% to 99.8% were observed in multiple studies.[20,21,23,28,30,32,34,40,44,50,52]

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Table 1 Included Studies About 3D-Printed Zirconia

Authors (year)	Study type	Manufacturing technology	Composition	Dental applications	Properties	Outcomes
Silva et al[17] (2011)	In vitro	RC	3Y-TZP (47 vol%)	Not specified	Morphology	Surface with "stair stepped" appearance, drying issues (eg, cracks) observed
Özkol et al[18] (2012)	In vitro	MJ and SC	MJ: 3Y-TZP (40 vol%), SC: 3Y-TZP (40 vol%)	Bridges	Density, microstructure, flexural strength	MJ samples have a relative density of > 96% TD, MJ samples revealed a smooth surface without "stair steps" effect and drying or sintering cracks, MJ samples presented higher flexural strength (≈ 843 MPa) than SC samples (≈ 684 MPa)
Jiang et al[19] (2014)	In vitro	MPSS (similar to DLP)	3Y-TZP	Copings	Shrinkage, density, microstructure, hardness, flexural strength	Shrinkage of 23.5%, density 98% TD, no delamination, hardness 1,328 HV, and flexural strength of 539 MPa was evident in the sample
Lee et al[20] (2015)	In vitro	3DSP (similar to DLP)	3Y-TZP	Implants	Shrinkage, density, microstructure, hardness, flexural strength	Shrinkage 32%, density 98.2% TD, no delamination, hardness 1,556 HV, flexural strength 542 MPa
Osman et al[21] (2017)	In vitro	DLP	3Y-TZP	Implants	Density, microstructure, surface roughness, flexural strength	Relative density 99.8% TD, varying surface roughness, flexural strength based on print orientation
Zhang et al[22] (2017)	In vitro	3D gel deposition and cold isostatic pressing	SG and CZ	Not specified	Microstructure, fracture force before and after fatigue tests	SG had higher fracture force than CZ, SG and CZ withstood occlusal forces
Lian et al[23] (2018)	In vitro	SLA	3Y-TZP	Bridges	Density, surface roughness, hardness, flexural strength	Relative density 98.58% TD, surface roughness 2.06 µm, hardness 1,398 HV, flexural strength 200.14 MPa
Hsu et al[24] (2019)	In vitro	3DSP (similar to DLP) and SM	3DSP: YSZ (75 wt%–34.5 vol%) SM: YSZ	Crowns	Marginal adaptation, hardness, flexural strength	Marginal adaptation higher for 3DSP, hardness higher for SM, 3DSP flexural strength 716.76 MPa
Jang et al[25] (2019)	In vitro	DLP	3Y-TZP (48–58 vol%)	Not specified	Density, cure depth, microstructure, flexural strength	Density increased with increase ZrO_2 vol%, cure depth decreased, flexural strength increased
Li et al[26] (2019)	In vitro	SLA	3Y-TZP	Bridges and implants	Microstructure	Cracks and pores observed on the surface
Li et al[27] (2019)	In vitro	SLA	ZrO_2 (45 vol%)	Crowns	Accuracy, flexural strength	Internal fit not ideal for clinical application, flexural strength 812 ± 128 MPa
Lian et al[28] (2019)	In vitro	MPSL (similar to DLP)	YSZ (40 vol%)	Crowns	Density, flexural strength	Relative density 99.3% TD and flexural strength 541 MPa
Branco et al[29] (2020)	In vitro	RC and SM	RC: 3Y-TZP (40 vol%) SM: 3Y-TZP	Crowns	Density, hardness, fracture toughness, cusp and prosthesis wear	RC samples' relative density of 98.3% TD, RC samples presented lower hardness (≈ 1,100 HV) and fracture toughness (≈ 4 MPa.m½) than SM samples (≈ 1,400 HV and ≈ 5.2 MPa*m½, respectively); no wear found on both RC and SM samples, RC samples induced lower cusp wear
Kim et al[30] (2020)	In vitro	DLP	4Y-PSZ (50 vol%)	Crowns	Density, microstructure, flexural strength	Relative density 99.4% TD, no visible interfaces, flexural strength 831 MPa
Li et al[31] (2020)	In vitro	DLP	3Y-TZP (40 vol%)	Crowns	Microstructure, hardness	Particles evenly distributed in cured resin matrix, hardness 1,038 HV

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Table 1 Included Studies About 3D-Printed Zirconia *(cont)*

Authors (year)	Study type	Manufacturing technology	Composition	Dental applications	Properties	Outcomes
Lu et al[32] (2020)	In vitro	DLP and SM	DLP: Y-TZP (58 vol%) SM: Y-TZP	Not specified	Density, microstructure, fracture toughness, flexural strength	DLP relative density 99% TD, similar microstructure, fracture toughness 5.4 MPa*m½, flexural strength 737.4 MPa between 2 samples
Marsico et al[33] (2020)	In vitro	DLP	5Y-PSZ	Not specified	Density, hardness, fracture toughness, flexural strength	Relative density 99.3% TD, other properties except hardness are dependent on printing orientations
Wu et al[34] (2020)	In vitro	DLP	ATZ (38.5 vol% ZrO_2)	Implants	Density, hardness, fracture toughness, ageing rate	Relative density 98.11% TD, hardness 1,290 HV, fracture toughness 6.42 MPa*m½ and samples displayed reduced rate of aging
Yu et al[35] (2020)	In vitro	RC	3Y-TZP (60 vol%)	Not specified	Density, hardness, fracture toughness, flexural strength	Relative density of 98.1% TD, hardness of 1,175 HV, fracture toughness of 2.63 MPa*m½, and flexural strength of 488.96 Mpa was evident
Shi et al[36] (2020)	In vitro	MJ	3Y-TZP (55 vol%)	Crowns	Density, hardness	Relative density of 98.5% TD and hardness of 1,468 HV was evident in the sample
Rodrigues et al[37] (2020)	In vitro	RC and SC	RC: 3Y-TZP (80 wt%-61.3 vol%) SC: 3Y-TZP (80 wt%-44.5 vol%)	Not specified	Density, hardness, fracture toughness	RC: Relative density ≥ 97%TD, Vickers hardness of 1,485 ± 32 HV, fracture toughness of 4.11 ± 0.09 MPa*m½; SC: Relative density ≥ 97% TD, Vickers hardness of 1,397 ± 27 HV, fracture toughness of 3.84 ± 0.21 MPa*m½
Fayazfar et al[38] (2020)	In vitro	MJ	5Y-TPZ (62.3 wt%-22.5 vol%)	Crowns	Density, hardness, fracture toughness	Relative density of 99.5% TD, hardness of 1,516 HV, and fracture toughness of 5.62 MPa*m½ were evident in the sample
Nakai et al[39] (2021)	In vitro	SLA and SM	SLA: 3Y-TZP and ATZ SM: 3Y-TZP	Implants	Microstructure and flexural strength	Microstructure and flexural strength of SLA samples were similar to SM samples
Chen et al[40] (2021)	In vitro	SLA	80 wt% 3Y-TZP + 20 wt% Al_2O_3	Implants	Density, hardness, fracture toughness	Relative density of 99.09% TD, hardness of 1,699 HV and fracture toughness of 6.88 MPa*m½ were observed
Coppola et al[41] (2021)	In vitro	DLP	Alumina-zirconia composites	Not specified	Microstructure, hardness, flexural strength, elastic modulus	Homogeneous microstructure, varying hardness 1,530–2141 HV, flexural strength 415–843 MPa, and elastic modulus 188–318 GPa
Lerner et al[42] (2021)	In vitro	DLP	3-TZP	Crowns	Trueness, precision	Higher trueness in SM crowns than DLP, similar precision for DLP and SM
Li et al[43] (2021)	In vitro	SLA and SM	SLA: 3Y-TZP (47 vol%) SM: PSZ	Crowns	Finish line designs evaluation	SLA crowns had margins of rounded line angles with minor flaws whereas SM crowns had margins of sharp line angles
Mei et al[44] (2021)	In vitro	DLP and SM	DLP: Y-TZP (58 vol%) SM: Y-TZP	Not specified	Density, microstructure, hardness, fracture toughness	Relative density 99% TD, similar microstructure, and fracture toughness between DLAP and SM samples but hardness for SM samples greater than DLP
Revilla-León et al[45] (2021)	In vitro	SLA	3Y-TZP	Crowns	Microstructure, fracture load, flexural strength, flexural modulus	SLA had varying properties based on porosity

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Table 1 Included Studies About 3D-Printed Zirconia *(cont)*

Authors (year)	Study type	Manufacturing technology	Composition	Dental applications	Properties	Outcomes
Tan et al[46] (2021)	In vitro	DLP and SM	DLP: 3Y-TZP (58 vol%) SM: 3Y-TZP	Implant abutment	Effect of accelerated aging on physical and biologic properties	Aging had no effect on cellular behavior but both zirconia type showed comparable biologic performance before and after aging
Willems et al[47] (2021)	In vitro	MJ	3Y-TZP (45 wt%-12.5 vol%)	Not specified	Density, microstructure, hardness, fracture toughness, elastic modulus	Density of 99.7% TD, hardness (≈ 1,285 HV) and fracture toughness (≈ 3.85 MPa*m½) independent of printing direction, flexural strength of 1,004 ± 138 MPa for samples printed in 0 degree orientation, elastic modulus higher when printed in 45 degree orientation (209 ± 5 MPa) than in 0 degree orientation (206 ± 5 MPa)
Xiang et al[48] (2021)	In vitro	SLA and SM	SLA: YSZ (84 wt%–48 vol%) SM: ZrO_2	Not specified	Dimensional accuracy, translucency, mechanical properties, microstructure	SLA samples demonstrated dimensional accuracy, translucency, and mechanical properties that varied with different build orientations; also, stress and weak bond strength were evident between layers of SLA samples
Zhai et al[49] (2021)	In vitro	SLA, DLP, and SM	SLA: ZrO_2 (50 vol%), DLP: ZrO_2, SM: ZrO_2	Not specified	Phase composition, microstructure, flexural strength, before and after aging	M-phase content increased with aging time, SLA samples showed grain pullout, DLP samples showed zirconia grain fragments, SM samples had no visible defects, SLA samples had the highest flexural strength
Zhao et al[50] (2021)	In vitro	DLP	5Y-PSZ (78 wt%–38.5% vol%)	Implant abutment	Density, hardness, flexural strength	Relative density 99.48% TD, varying properties based on printing orientation
Abualsaud et al[51] (2022)	In vitro	SLA	3Y-TZP	Crowns	Trueness, precision	SLA crowns had the best occlusal trueness, similar internal fit and higher precision were observed between SM and SLA samples
Coppola et al[52] (2022)	In vitro	DLP	3Y-TZP (40.5 vol%)	Implants	Density, microstructure, flexural strength, elastic modulus, hardness	Relative density 99.2% TD, properties influenced by printing direction and zirconia content
Jang et al[53] (2022)	In vitro	DLP	3Y-TZP (52, 54, 56 vol%)	Not specified	Density, strength, hardness	Density increased with ZrO_2 vol%, silane coupling agent improved strength and hardness
Kim et al[54] (2022)	In vitro	DLP, SLA, and SM	SLA: 3Y-TZP, DLP: 3Y-TZP, SM: 4Y-PSZ/5Y-PSZ	Crowns	Trueness, antagonist wear, microstructure	Similar trueness of intaglio crown surfaces, similar volume loss of antagonist teeth for DLP, SLA, and SM samples
Lüchtenborg et al[55] (2022)	In vitro	SLA, MJ, DLP, and SM	SLA: 3Y-TZP, MJ: 3Y-TZP, DLP1: 3Y-TZP, DLP2: 3Y-TZP, SM: 3Y-TZP	Not specified	Accuracy, surface deviation	SM led to the most accurate samples, DLP1 had the least accurate samples
Meng et al[56] (2022)	In vitro	DLP	3Y-TZP (40 vol%)	Crowns	Dimensional accuracy	DLP crowns had good internal fit and marginal adaptation
Moon et al[57] (2022)	In vitro	DLP and SM	DLP: ZrO_2 SM: ZrO_2	Copings	Shrinkage, accuracy, bond strength	DLP led to higher thermal shrinkage and lowest accuracy, higher bond strength. But adhesive failure was seen between porcelain and zirconia

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Table 1 Included Studies About 3D-Printed Zirconia *(cont)*

Authors (year)	Study type	Manufacturing technology	Composition	Dental applications	Properties	Outcomes
Revilla-León et al[58] (2022)	In vitro	SLA	3Y-TZP	Crowns	Microstructure, flexural strength	SLA had a smooth surface with no cracks, higher flexural strength
Gatto et al[59] (2022)	In vitro	DLP and SM	DLP: 3Y-TZP SM: 3Y-TZP	Not specified	Microstructure, tendency to deformation under compression	A few DLP samples failed, while all samples of SM did not fail under the load cell limit; compared to SM samples, DLP samples had lower tendency to deform under compression
Santos et al[60] (2022)	In vitro	RC	5Y-PSZ (42 vol%)	Not specified	Density, hardness, fracture toughness, flexural strength	Relative density ≤ 94% TD, hardness of 1295 HV, fracture toughness of 3.91 MPa*m½ and flexural strength of 285 MPa was evident in the samples

5Y-PSZ = 5-mol% yttria partially stabilized zirconia; ATZ = alumina-toughened zirconia; CZ = conventional zirconia; MPSL = mask projection stereolithography; SG = self-glazed zirconia; SC = slip casting; SM = subtractive manufacturing or milling; YSZ = yttria-stabilized zirconia.

Table 2 Included Studies About 3D-Printed LS_2

Authors (year)	Study type	Manufacturing technology	Dental application	Properties	Results
Unkovskiy et al[61] (2022)	Clinical report	LCM	Veneers	Marginal adaptation	Adequate esthetics and a sufficient marginal fit of 100 µm
Schweiger et al[62] (2024)	Clinical report	LCM	Veneers	Total layer thickness	LCM technology enables the production of ultra-thin LS_2 veneers with layer thicknesses of down to 0.2 mm
Baumgartner et al[63] (2020)	In vitro	SLA	Not specified	Mexhanical properties and rheological behavior	The processing of LS_2 via an 3D-printing technology offers highly dense (> 99%), full ceramic parts that meet the requirements for their use as dental restorations
Marsico et al[64] (2022)	In vitro	SLA	Not specified	Flexure strength, Weibull modulus, and elastic modulus	3D-printed LS_2 materials can achieve the mechanical properties of materials produced by traditionally processing
Abreu et al[8] (2023)	In vitro	RC	Not specified	Mechanical properties, porosity	Mechanical properties of 3D-printed LS_2 were weak and porosity was high compared to milled ones
Kim et al[65] (2023)	In vitro	DLP	Not specified	Shrinkage rate, Vickers hardness, and translucency	LS_2 is a promising material for the 3D printing of dental prostheses; however, 3D printing of ceramics using photopolymerization involves many steps, and different factors at each stage must be determined carefully to result in an ideal manufactured product

In several studies, the flexural strength of 3D-printed zirconia was measured to be in the range of 200 to 831 MPa,[19–21,23–25,27,28,30,32,41,45,49,50,52] which is lower than the range of 900 to 1,200 MPa observed in the zirconia manufactured via conventional manufacturing techniques.[47,66,67] Nakai et al[39] reported that SLA 3D-printed samples had flexural strength comparable to those fabricated from subtractive manufacturing techniques. On the contrary, several studies reported high flexural strength within the range of 943 to 1,519 MPa using the DLP or SLA technique.[21,26,32,49] This was attributed to improved slurry composition, debinding, and sintering processes.[4] Hardness was between 1,038 to 1,556 HV, and fracture toughness was between 3 to 6 MPa, comparable to conventional manufacturing methods to produce zirconia.[20,24,28,31,32,34,41,44,50] Unlike SLA and DLP techniques, studies on the fabrication of 3D-printed zirconia using RC and MJ are limited. However, from the available literature, the theoretic density of RC and MJ samples were reported to be 94% to 98%[35,37,60] and 96% to 99%,[18,36,38,47] respectively. Moreover, other mechanical properties were similar to DLP and SLA techniques.[4]

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Defects

Defects in 3D-printed zirconia, such as cracks, pores, and fractures, are well-documented. Li et al[26] observed cracks and porosities on the surfaces of SLA 3D-printed zirconia. Osman[21] found cracks and microporosities, while Marsico et al[33] noted fractures originating from layer lines. On the other hand, Revilla-León et al[58] saw no cracks or fractures but identified pits under a scanning electron microscope (SEM). Willems et al[47] reported cracks and surface porosities using MJ from a slurry with low solid content. Gatto et al[59] reported small pores of about 3 μm on the surface of DLP 3D-printed samples. However, DLP samples had a low tendency to deform under compression. Jang et al[25] mentioned that increased zirconia during processing improved crack resistance. Xiang et al[48] linked surface defects to higher restoration failure rates. Zhang et al[22] found that self-glazed zirconia restorations fabricated using a 3D-printed gel deposition approach had a higher fracture force than traditional ones, attributing this to fewer grains and voids.

Stair-step artifacts

The stair-stepping effect, a common flaw in 3D printing, is marked by visible layering in the final object. To ensure esthetic quality, it is crucial to minimize the appearance of these layers. Revilla-León et al[58] achieved a layer strand texture with a 5 to 10 μm smooth depression between layers using SLA technology. Silva et al[17] encountered a stair-stepped look in dental prosthesis manufacturing with the RC technology. In contrast, Özkol et al[18] reported a smooth, stair-step-free surface using MJ technology. Li et al[31] and Kim et al[30] found that the layered structure fades post-sintering. Additionally, applying glass veneers over zirconia restorations can enhance esthetics by concealing any layered appearance.

Polymerization depth

Among several 3D-printing techniques, DLP and SLA use UV or laser light for polymerization or solidification of the final 3D-printed objects. Because zirconia has a high refractive index, it tends to cause light scattering and reduce the polymerization depth.[25,31,69,70] Similarly, the presence of zirconia with small particle sizes or a high concentration of solids in the suspension further decreases the polymerization depth. This effect results in excess ceramic material loosely attached to the final 3D-printed objects.[25,71] Furthermore, in DLP or SLA 3D-printing processes, maintaining the homogeneity of the ceramic suspension, which consists of ceramic particles and a photosensitive resin, is crucial for a duration ranging from hours to days. Achieving this homogeneity is possible through various procedures, including adding dispersants and other additives to the suspension, coating of particles, ultrasonication, vacuum drying, and acid treatment.[4]

Bond strength

A study by Moon et al[57] compared the bond strength of porcelain to zirconia produced through both 3D-printing and milling methods. The authors concluded that the bond strength between porcelain and 3D-printed zirconia was higher than that of milled zirconia. This conclusion suggests a higher potential for 3D-printed zirconia in dentistry.

Wear behavior of zirconia prostheses and antagonists

Kim et al[54] used 3D-printed 3 mol% yttria-stabilized tetragonal zirconia polycrystal (3Y-TZP) and conducted a wear simulation using human molar teeth as antagonists. The authors reported a wear volume loss of 2.06 ± 1.24 mm³ for DLP and 1.74 ± 1.20 mm³ for SLA techniques. These values were in line with those of samples manufactured through milling techniques. Similarly, Branco et al[29] performed a chewing simulation test on zirconia samples manufactured using 3D-printing and milling methods against natural teeth. No wear was reported on the samples from either technique. However, wear was detected on the antagonist teeth caused by the zirconia samples from both fabrication techniques, with milled samples showing significantly more wear. Additionally, after both types of zirconia samples were glazed, they underwent further chewing simulation tests against human teeth. The authors noted that the glazed samples produced with 3D printing caused less wear on the antagonist cusps than those produced by the milling method.[29]

Printing orientation

Printing orientation is crucial during 3D-printing because it affects the quality of the final objects. For example, Xiang et al[48] found that in SLA 3D printing of zirconia, samples manufactured in a vertical orientation exhibited superior density and optical properties compared to those in a horizontal orientation. However, the horizontal orientation was associated with greater precision and mechanical properties. Similarly, Coppola et al[52] used the DLP technique for 3D-printing zirconia and observed that samples printed in a vertical orientation demonstrated enhanced flexural strength. Zhao et al[50] corroborated these findings concerning flexural strength. In contrast, Marsico et al,[33] using the DLP technique at 0-degree, 45-degree, and 90-degree orientations, discovered that samples at a 45-degree orientation exhibited the highest resistance to indentation fractures. Osman et al[21] reported that zirconia samples 3D printed using the DLP technique had the highest flexural strength at a 0-degree orientation and the lowest at a 45-degree orientation.

Marginal and internal fit

The success of dental restorations, produced using either 3D-printing or milling manufacturing methods, largely

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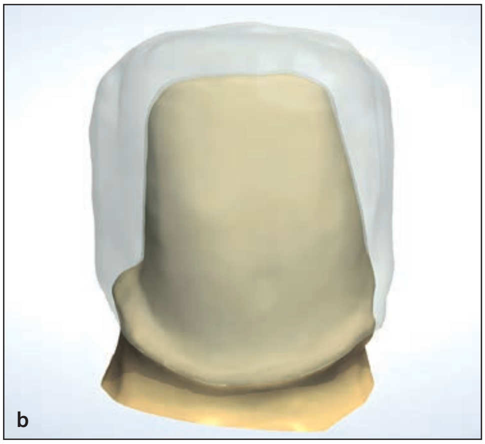

Figs 2a and 2b *(a)* Preoperative patient presentation. *(b)* Preliminary design of zirconia substructure. →

depends on their internal adaptation and marginal fit.[51] Accuracy in this context is defined by trueness and precision. Ensuring high levels of both during the fabrication process is crucial for achieving a proper fit and favorable biologic response.[4] Lerner et al[42] evaluated the accuracy of zirconia crowns manufactured using DLP 3D-printing vs milling techniques. They found that milled crowns had greater trueness, although the precision of the marginal fits was similar for both methods. Moon et al[57] fabricated zirconia crowns using SLA 3D-printing and milling, observing comparable trueness across all parameters—including the external surface, intaglio surface, marginal area, and occlusal surface—for both methods. Kim et al[54] also fabricated crowns using DLP and SLA 3D-printing in addition to milling and noted consistent accuracy in the inner surface area across all techniques but observed variations in trueness in the occlusal, marginal, and axial areas. Lüchtenborg et al[55] compared the precision of fixed dental prostheses made using SLA, DLP, MJ, and milling, determining that milling provided the most accurate results. Meng et al[56] discovered that milled zirconia crowns' internal fit and marginal adaptation were superior to those made using DLP 3D-printing. In contrast, Hsu et al[24] reported better marginal adaptation for premolar crowns made with DLP 3D printing than milling, though still within clinically acceptable ranges. Li et al,[27] however, found the SLA technique unsuitable for dental applications due to a cement space of approximately 170 µm in the marginal area.

Clinical applications in dentistry

In medicine, ceramics are mainly used to fabricate hip implants, knee implants, tissue engineering, and scaffolds. Meanwhile, in dentistry, zirconia-based ceramics are primarily used for dental restorations, dental implants, bone generation, and bone tissue engineering.

Dental prostheses

Zirconia's excellent flexural strength (900 MPa), biocompatibility, and stability in body fluids make it popular for dental crowns and bridges.[72] Ebert et al[73] achieved high-density (96.9%) zirconia prostheses with 763 MPa strength and 6.7 MPa fracture toughness using direct inkjet printing (DIP), although some samples showed reduced strength due to nozzle clogging. Yves-Christian et al[74] used the SLM technique to manufacture Al_2O_3-ZrO_2 specimens, obtaining crack-free, dense components with over 500 MPa flexural strength, suitable for medical and dental use. Lian et al[28] enhanced the bending strength of customized zirconia prostheses with mask projection SLA, surpassing human dentin. Özkol et al[18] explored the DIP technique for 3Y-TZP fixed dental prostheses, achieving smooth, crack-free surfaces and notable physical properties, with tensile strength between 250 to 350 MPa and 843 MPa flexural strength.

This clinical case report describes the application of 3D-printed zirconia crowns for restoring maxillary central incisors. The patient initially presented with interim crowns (Fig 2a) on these teeth. These crowns were removed, and the abutment teeth underneath were cleaned and re-prepared. A definitive impression was taken, and the Type 5 dental stone was used to pour a definitive cast. This cast was digitized using a laboratory scanner (E4, 3Shape). The zirconia substructures' preliminary design was rendered using a dental CAD/CAM software (Dental System, 3Shape; Fig 2b), with additional internal retentive structures incorporated using an industrial CAD/CAM program (Blender, Blender Studio; Fig 2c). The substructures were 3D printed with an advance customized jetting (ACJ) ceramic 3D printer (PGJ180, Thales Medtech), using a 3Y-TZP ceramic nanoparticle suspension (Fig 2d). After printing, the support material was removed in a water bath, and the zirconia substructures were sintered in a high-temperature furnace to

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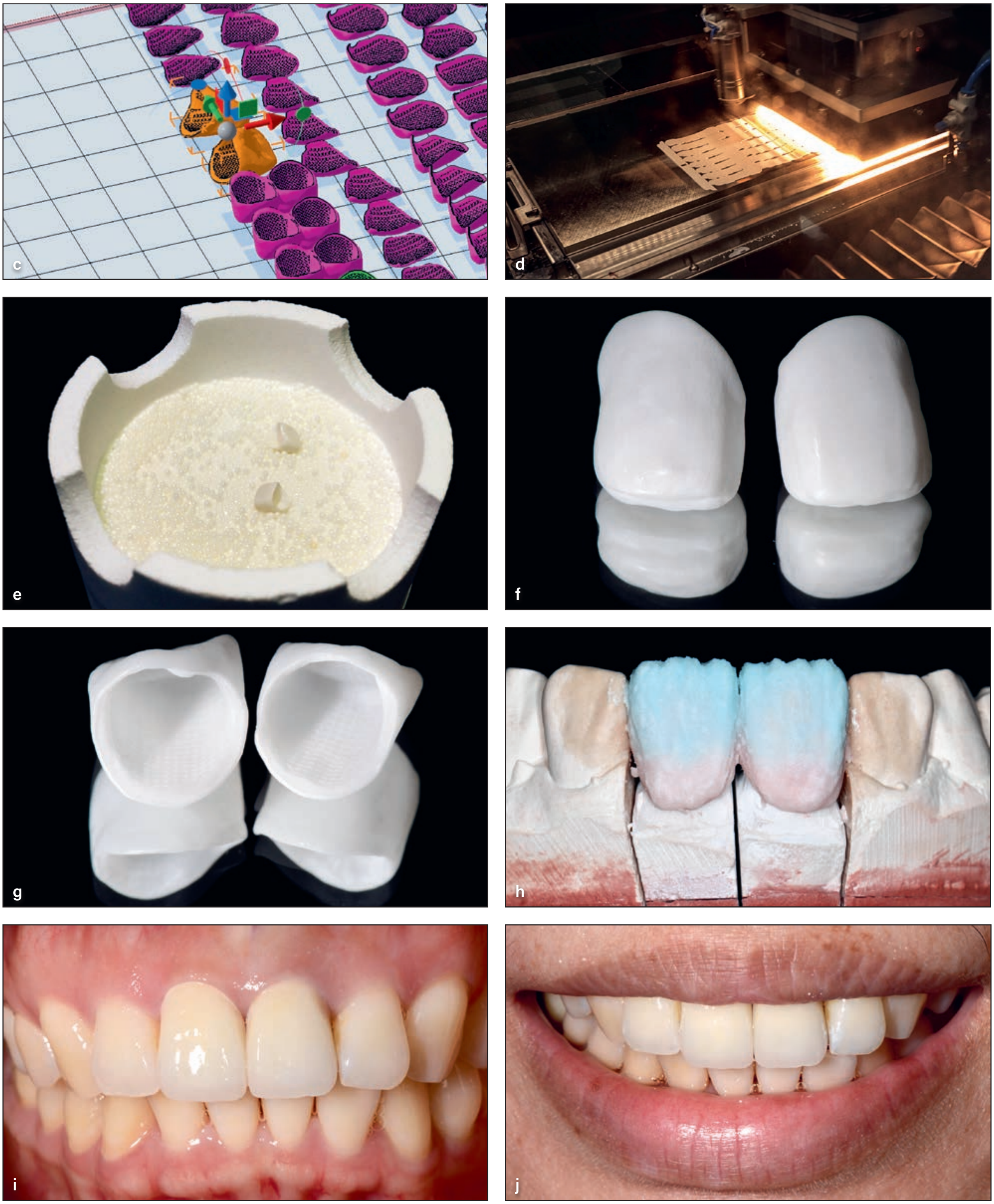

Figs 2c to 2j *(c)* Additional internal support structures were incorporated in the design before 3D printing. *(d)* An ACJ ceramic 3D printer was used to manufacture the zirconia substructures. *(e)* Zirconia substructures were sintered in a high-temperature furnace. *(f)* Outer surface view of sintered zirconia substructures. *(g)* Intaglio view of sintered zirconia substructures. *(h)* Feldspathic porcelain was layered onto the zirconia substructures. *(i and j)* Definitive zirconia crowns in situ after cementation.

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Fig 3 *(a)* Preoperative patient presentation. *(b)* Feldspathic porcelain was layered onto the zirconia substructure. *(c)* Definitive zirconia crowns in situ after cementation.

eliminate any remaining bonding agents (Figs 2e to 2g). It's important to recognize the retentive features incorporated on the intaglio surfaces of zirconia substructures (Fig 2g). Such features cannot be created using milling technologies. Their use alongside 3D-printed zirconia could enhance the retention of dental restorations significantly. Feldspathic porcelain was then layered onto the zirconia substructures to construct the definitive crowns with the necessary morphologic contours and optical properties (Fig 2h). These crowns were cemented using a self-adhesive, dual-curing resin cement (RelyX Unicem 2 Automix, 3M; Figs 2i and 2j). Another clinical case report describes the application of 3D-printed zirconia crowns for restoring the maxillary left central incisor with a metal cast post and core (Fig 3a). Similar clinical and laboratory processes were used to obtain definitive impression, cast, and 3D-printed zirconia substructure with additional internal retentive structures. Feldspathic porcelain was then layered onto the zirconia substructures (Fig 3b). The crown was cemented using a self-adhesive, dual-curing resin cement (3M; Fig 3c).

Dental implants

Presently, zirconia can be used to fabricate dental implants (Fig 4). Research suggests that zirconia implants result in faster and stronger integration with the underlying bone due to the chemical bond formed between the two, unlike titanium implants, which rely on mechanical interlocking.[15] In addition, zirconia implants lower the risk of peri-implantitis, which is common with metal-based implants.[75] One study used DLP 3D-printed zirconia root analog implants (RAI) based on CBCT data and evaluated its accuracy.[76] That study demonstrated that RAI had a 7% more surface area, and 46% of the RAI showed a greater divergence for surface area, surpassing the 0.1 mm threshold. In addition, the authors concluded that DLP technology could fabricate customized zirconia implants with adequate dimensional stability and flexural strength. Another study by Osman et al[21] harnessed the DLP technique to 3D print custom zirconia implants. The authors concluded that the DLP method could produce customized zirconia dental implants with satisfactory dimensional accuracy and mechanical strength like conventionally manufactured ceramics. Another study by Lee et al[20] developed a 3D slurry printing system (3DSP) and implemented a two-stage sintering process. This approach successfully produced zirconia dental implants, with the sintered parts exhibiting an average flexural strength of 539.1 MPa and a microhardness of 1,556 HV.

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Fig 4 Experimental one-piece 3D-printed zirconia implant. The substructure was 3D printed with an ACJ ceramic 3D printer (PGJ180; Thales Medtech) using a 3Y-TZP ceramic nanoparticle suspension.

Bone generation and bone tissue engineering

Traditional biomaterials cannot replicate a complex extracellular environment for maintaining cell viability and function. As a solution, 3D porous scaffolds with specific cells for bone regeneration have been explored. Zirconia's similarities to human bones in elastic modulus, fracture resistance, and osseointegration properties make it a focus of research in bone tissue engineering. Furthermore, 3D-printed zirconia-based scaffolds are gaining popularity due to their superior mechanical and biologic properties.[72] Li et al[31] fabricated zirconia scaffolds using the direct-ink-writing (DIW) technique. In this process, water-based zirconia ink with a 70% solid content was deposited layer by layer to achieve the desired structure. The authors reported that the compressive strength of the 3D-printed zirconia scaffold was better than the hydroxyapatite (HA) one. The proliferation of HCT116 cells in the vicinity of the 3D-printed zirconia scaffolds was also observed microscopically. Thus, the DIW technique can be considered for 3D-printed zirconia-based scaffold production. Another study by Kocyło et al[77] fabricated 3D-printed zirconia scaffolds with 61 and 75.3 vol% porosity using the DIW method. The SEM results demonstrated uniformity in zirconia scaffolds with good control in thread and pore openings. Shuai et al,[78] in their study, fabricated a nano zirconia-reinforced calcium silicate ($CaSiO_3$) porous scaffold ($CaSiO_3$/nanozirconia scaffold) using the SLS technique. The authors concluded that the addition of nano-zirconia resulted in superior mechanical properties. However, an increase in the zirconia concentration beyond 30% affected the sintering process, which led to undesirable agglomeration and compromised scaffold material properties.

3D-Printed LS_2 Ceramics

Mechanical properties

The mechanical properties of 3D-printed LS_2 ceramics have been evaluated in a few studies. Abreu et al[8] conducted a study to evaluate the mechanical properties of 3D-printed LS_2 ceramic structures. In their research, disc-shaped samples of LS_2 were fabricated using milling and RC techniques. The mechanical properties of samples produced through these two techniques were assessed and compared using a biaxial flexural strength test and Vickers hardness test. The SEM was used to study the fracture origin and crack propagation from the fractographic images of the samples. The authors reported higher biaxial flexural strength and hardness values in milling group samples (325.09 ± 63.98 MPa and 5.63 ± 0.14 GPa, respectively) compared to RC groups (120.02 ± 33.91 MPa and 4.07 ± 0.30 GPa, respectively). The SEM study demonstrated higher porosity in 3D-printed samples compared to milled samples. Moreover, Fourier transform infrared spectroscopy (FTIR-ATR) and x-ray diffractometry (XRD) indicated a lower crystalline structure and a decrease in the formation of LS_2 ($Li_2O_5Si_2$) from lithium metasilicate ($Li_2O{-}SiO_2$), respectively, in the 3D-printed group. The authors concluded that samples fabricated using milling reported better mechanical properties than 3D-printed samples. However, 3D printing can be successfully used to fabricate $Li_2O_5Si_2$ ceramic structures using a 3D-printing technique.[79]

Another study by Kim et al[65] evaluated the 3D-printed dental prostheses fabricated using the sol-gel method. To overcome the decrease in the strength of 3D-printed samples after sintering, the sol-gel method was adopted to synthesize a pure form of LS_2. In their study, the 3D-printed LS_2 samples were manufactured using the DLP technique with different holding times (1, 3, 5, and 10 hours) and a sintering temperature of 1,325°C. The mechanical properties such as shrinkage rate, Vickers hardness, and translucency were analyzed for the samples and compared to those not 3D printed. The study's results revealed a significantly increased shrinkage rate to 7.06% and a significant decrease in translucency to 66.06% in 3D-printed LS_2 samples. The Vickers hardness value decreased significantly to 53.24% in the 3D-printed LS_2 samples but was comparable to heat-pressed or milled lithium silicate samples. The authors concluded that 3D-printed LS_2 can be considered a promising material for fabricating dental prostheses. However, the photopolymerization technique used for 3D-printing ceramics involves multiple steps, requiring thorough analysis to achieve ideal results.

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On the contrary, a study by Baumgartner et al[63] evaluated the accuracy, mechanical properties, and reproducibility of LS_2 dental restorations using a DLP 3D printer. Authors reported that definitive restorations encompassing veneers, crowns, and bridges exhibited a high density (> 99%), minimal porosities, flexural strength exceeding 400 MPa, superior translucency, and accuracy. Another study by Marsico et al[64] evaluated the mechanical properties of 3D-printed LS_2 structures fabricated using the DLP technique. Authors assessed and compared the impact of build orientation on the fracture resistance of 3D-printed structures with samples fabricated from traditional processes. Following sintering and post-processing, 3D-printed bars were acquired in three orientations (0 degrees, 45 degrees, and 90 degrees) relative to the build direction. Authors reported that at 0 degrees orientation, 3D-printed structures showed flexure strength, Weibull modulus, and elastic modulus of 313 MPa, 4.42, and 168 ± 3 GPa, respectively, similar to LS_2 structures fabricated from traditional processes. It was noted that unlike flexure strength, hardness, and fracture toughness were not dependent on build orientation. The authors concluded that 3D-printed LS_2 structures can attain mechanical properties similar to traditionally processed ones.

Clinical applications

Limited literature is available on the application of 3D-printed LS_2 restoration in clinical settings. A study by Unkovskiy et al[61] was the first to demonstrate the clinical application of 3D-printed LS_2 restorations with LCM technology in the anterior region. In their study, six veneers were fabricated using LCM technology to restore severely worn teeth in the mandibular anterior region and non-prep veneers for a diastema closure in the maxilla. The fabricated veneers were evaluated in terms of marginal fit. Additionally, the feasibility of 3D printing of non-prep LS_2 veneers with 0.1-mm thickness was also analyzed. The authors reported adequate esthetics and sufficient marginal fit (within 100 μm). Similarly, Schweiger et al[62] evaluated the feasibility of the LCM 3D-printed ultra-thin LS_2 veneers for the maxillary anterior region. The authors concluded that this technology demonstrates better economic, mechanical, and esthetic outcomes than existing technologies. Another study by Abreu et al[8] compared the efficacy of 3D-printed and milled LS_2 ceramic constructs. It concluded that the mechanical properties of 3D-printed structures were inferior and more porous than milled samples.

Challenges and Future Considerations for 3D-Printed Zirconia and LS_2

Zirconia and LS_2 ceramics, exceptional biomaterials in dentistry, face challenges in 3D-printing applications, especially when produced using 3D printing. A significant drawback arises when the size of the particles used in the process is inconsistent, leading to light scattering that affects the material's curing depth. This effect results in loosely attached ceramic particles in the final 3D-printed product.[4] Another challenge in 3D-printed zirconia is the presence of pores and cracks on the surface of 3D-printed objects. The layer-by-layer 3D-printing process increases the chances of cracks between layers and the formation of defects, ultimately leading to esthetically unsatisfactory final objects.[4] Additionally, a well-controlled debinding process is essential to fabricate crack-free ceramic structures.[65] Even inadequate holding time before removal of the binder can initiate cracks. Unlike human teeth, which exhibit transparency in light, 3D-printed zirconia appears opaque. 3D-printed LS_2 is relatively weaker than zirconia but offers unmatched high esthetics.[63] Furthermore, printing orientation is another vital factor to consider during 3D printing. The printing orientation significantly influences 3D-printed objects' quality, accuracy, surface roughness, translucency, and mechanical properties. The industrialization of 3D-printing with zirconia or LS_2 poses significant challenges due to the high cost and technical sensitivity involved in producing high-quality, accurately sized bioceramics compared to conventional methods. Meeting stringent safety standards further complicates certification and quality assessment.[5]

3D printing represents the future of the medical and dental fields, particularly in prosthetic rehabilitation within dentistry. To ensure the success of this method, creating awareness about 3D printing among trainees and dentists is crucial. Many dentists may hesitate to adopt this method due to its technical sensitivity and cost. Raising awareness through mass media and educational initiatives is essential for the sustainable future of 3D printing. For a more practical application of 3D-printed ceramics in dentistry, efforts should concentrate on reducing excessive costs associated with the machine, its raw materials, and the 3D-printing process itself. Research must be carried out to decrease production defects, such as porosities and cracks, and develop methods to enhance the strength of the final product while reducing surface shrinkage. Through further studies, solutions to these problems can be identified, ensuring a high success rate of 3D printing using zirconia, LS_2, or similar biomaterials. If these challenges are overcome, 3D-printed ceramic materials could become a chairside process in the future.

CONCLUSIONS

3D printing of zirconia-based and LS_2-based ceramic materials has a promising future in dentistry. These materials, known for their biocompatibility and esthetic qualities resembling natural teeth, have demonstrated

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high success rates. Various 3D-printing methods are available, with DLP and SLA showing improved final products. Despite limitations and challenges, such as defects, compromised mechanical strength, and esthetic concerns, ongoing research aims to enhance the quality and esthetics of 3D-printed ceramic materials. Efforts are also needed to reduce the high costs associated with this form of manufacturing. Currently, awareness among clinicians and technicians is inadequate and must be addressed. These challenges can be overcome by conducting studies focusing on reducing surface defects, improving dimensional stability, and enhancing the mechanical strength of 3D-printed prostheses while increasing awareness of this new technique.

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