

Review

Additive Manufacturing of Zirconia and Its Clinical Applications in Dentistry

Mengzhe Cai^{*1}, Bingjie Wang¹, Tingting Cao², Luyi Chai¹

¹Department of Stomatology, The Affiliated People's Hospital of Ningbo University (Ningbo Yinzhou People's Hospital), Ningbo, 315000, China

²Ningbo Haishu Stomatology Hospital, Ningbo, 315000, China

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Abstract

The emergence of additive manufacturing (AM) has catalyzed a paradigm shift in the fabrication of dental zirconia restorations, challenging the long-standing dominance of subtractive manufacturing (SM) methods. AM techniques – including vat photopolymerization, material extrusion, and binder jetting – offer unprecedented design freedom, near-net-shape fabrication, and reduced material waste, enabling the production of complex, patient-specific restorations. Unlike milling, which relies on standardized industrial blanks, AM defines material properties through a process-dependent workflow encompassing feedstock formulation, printing parameters, debinding, and sintering. Recent studies demonstrate that optimized AM zirconia can approach or match SM in flexural strength, hardness, and phase composition, though significant variability persists. Microstructural anisotropy, residual porosity, and interlayer defects remain critical factors limiting fatigue resistance and long-term reliability. Dimensional accuracy, while statistically inferior to SM, frequently meets clinical acceptability thresholds, and translucency improvements are rapidly closing the aesthetic gap. Biocompatibility assessments confirm the safety of fully sintered AM zirconia, though rigorous post-processing protocols are essential to eliminate residual cytotoxic agents. Clinical applications currently focus on single-unit crowns, with expanding potential for more complex prostheses as process standardization and quality control improve. Despite substantial progress, the lack of randomized clinical trials and standardized manufacturing pathways remains a central barrier to widespread adoption. This review critically synthesizes the evolving evidence on AM zirconia, providing a comprehensive framework for evaluating its performance, challenges, and translational potential in restorative dentistry.

Keywords: Microstructure, fatigue resistance, dimensional accuracy, translucency, biocompatibility

I. Introduction

Zirconia has established an unparalleled position in modern restorative dentistry, emerging as the material of choice for a vast array of clinical indications due to its superior mechanical strength, excellent biocompatibility, and progressively improving aesthetic properties¹. Specifically, 3 mol% yttria-stabilized tetragonal zirconia polycrystals (3Y-TZP) have become the benchmark for high-strength restorations, enabling the fabrication of durable crowns, multi-unit fixed dental prostheses (FDPs), and implant abutments that can withstand the demanding occlusal forces of the posterior dentition². The widespread adoption of zirconia was catalyzed by the maturation of digital dentistry, particularly computer-aided design/computer-aided manufacturing (CAD/CAM) systems that employ a subtractive manufacturing (SM) or milling approach. This technology transformed dental fabrication, offering unprecedented precision and reproducibility compared to traditional

casting techniques³. However, after decades of clinical dominance, the inherent limitations of the SM paradigm are becoming increasingly apparent, creating a fertile ground for disruptive innovation.

The subtractive manufacturing workflow, while reliable, is fundamentally a process of material removal, which introduces significant inefficiencies and constraints. A primary critique is the substantial material wastage; studies indicate that fabricating a single restoration can involve milling away as much as 80 % to 90 % of the initial zirconia blank, contributing to higher production costs and generating considerable environmental waste⁴. Beyond economic concerns, SM imposes significant geometric limitations. The physical dimensions of the milling burs, including their diameter and length, restrict the ability to create truly complex and anatomically refined structures. This is particularly evident in the challenges of accurately machining intricate occlusal fissure morphology, fine marginal details, and thin-walled copings, potentially compromising the final restoration's

* Corresponding author: cai_mengzhe@163.com

fit and function⁵. Furthermore, the mechanical process of milling, whether in the pre-sintered (“soft”) or fully sintered (“hard”) state, introduces the risk of surface and subsurface damage. Tool wear can lead to a gradual loss of precision, while the machining process itself can induce microcracks and residual stresses that may act as initiation sites for fracture, thereby compromising the long-term clinical performance and reliability of the zirconia restoration⁶.

In direct response to these shortcomings, additive manufacturing (AM), colloquially known as 3D printing, has emerged as a compelling and potentially revolutionary alternative paradigm for zirconia fabrication. Unlike the top-down approach of SM, AM is a bottom-up, layer-by-layer process that constructs objects directly from a digital file. This methodology offers profound theoretical advantages that directly address the primary weaknesses of milling. The ability to create near-net-shape components virtually eliminates material waste, offering a more sustainable and cost-effective manufacturing route⁷. Perhaps more significantly, AM unshackles dental restorations from the geometric constraints of milling tools, enabling unparalleled design freedom. This allows for the fabrication of highly complex structures, such as customized implant abutments with intricate internal features, frameworks with optimized stress distribution, or scaffolds with controlled porosity designed to enhance osseointegration⁸.

However, the transition from SM to AM is not merely an exchange of one fabrication tool for another; it represents a fundamental shift in how the final material is created. In an SM workflow, the material properties are largely predetermined by the industrially manufactured, quality-controlled zirconia blank. The milling process shapes this pre-existing material. In contrast, AM is a process where the material and the object are created simultaneously. The final properties of an AM zirconia part are not inherent to the starting powder alone but are a direct output of the entire, multi-stage process chain: the formulation of the ceramic slurry or filament, the specific printing parameters such as layer thickness and build orientation, and, most critically, the complex thermal post-processing steps of debinding and sintering⁹. This reality, where the process itself defines the material's final microstructure and performance, lies at the heart of the current debate. The central question is not simply whether AM zirconia is equivalent to SM zirconia, but whether an entire AM workflow can be controlled with sufficient precision and reliability to consistently produce dental restorations that meet or exceed the high standards set by decades of clinical success with milled zirconia.

Despite its immense promise, the clinical adoption of AM zirconia is currently impeded by significant and valid controversies surrounding its mechanical reliability, microstructural integrity, dimensional accuracy, and long-term stability when compared to the well-established SM gold standard¹⁰. The published literature presents a landscape of conflicting results, with some studies demonstrating comparable performance while others highlight significant deficiencies. Therefore, the

objective of this review is to critically analyze the current body of scientific evidence on the additive manufacturing of zirconia for dental applications. This paper will dissect the core arguments surrounding the various AM technologies, examine the conflicting findings on material properties, and identify the key technological hurdles and research gaps that must be surmounted before AM zirconia can transition from a promising experimental technology to a trusted standard of clinical care. To frame the debate beyond a tool swap, **Fig. 1** contrasts subtractive and additive zirconia along three value axes – material utilization, geometric freedom, and damage risk. It emphasizes that AM is a process-defined material: properties emerge from slurry/ink formulation, build parameters, and thermal post-processing, not from an industrial blank. This systems view motivates the review's evidence-based assessment of whether end-to-end AM control can meet clinical benchmarks established by SM zirconia.

II. Additive Manufacturing Technologies for Zirconia Fabrication

The fabrication of dense, high-strength zirconia components via additive manufacturing is a complex undertaking that has given rise to several distinct technological approaches. Each method presents a unique set of advantages, challenges, and process-specific considerations that directly influence the quality and properties of the final restoration. A comprehensive understanding of these technologies is essential to appreciate the nuances of the current debate surrounding AM zirconia. The processes can be broadly categorized based on their fundamental principle: vat photopolymerization, material extrusion, and binder jetting. Crucially, all these methods share a common multi-stage workflow that involves printing a “green” composite part, followed by essential thermal post-processing to remove binders and densify the ceramic.

(1) Vat Photopolymerization (VPP) technologies (SLA and DLP)

Vat photopolymerization (VPP) technologies are currently the most investigated and arguably the most promising methods for producing high-precision dental zirconia restorations. These processes work by selectively curing successive layers of a liquid photopolymer resin heavily loaded with ceramic particles, known as a slurry¹¹. The two primary VPP techniques are stereolithography (SLA) and digital light processing (DLP). SLA utilizes a high-precision, point-source ultraviolet laser, guided by galvanometers, to trace the cross-section of each layer, polymerizing the slurry point-by-point. In contrast, DLP employs a digital projector, often using a digital micromirror device (DMD), to project an image of the entire layer at once, curing it in a single exposure¹². This fundamental difference results in a key trade-off: DLP offers significantly higher build speeds, making it attractive for clinical and laboratory settings, while the continuous tracing motion of SLA can result in parts with greater isotropy and potentially lower internal stresses¹³.

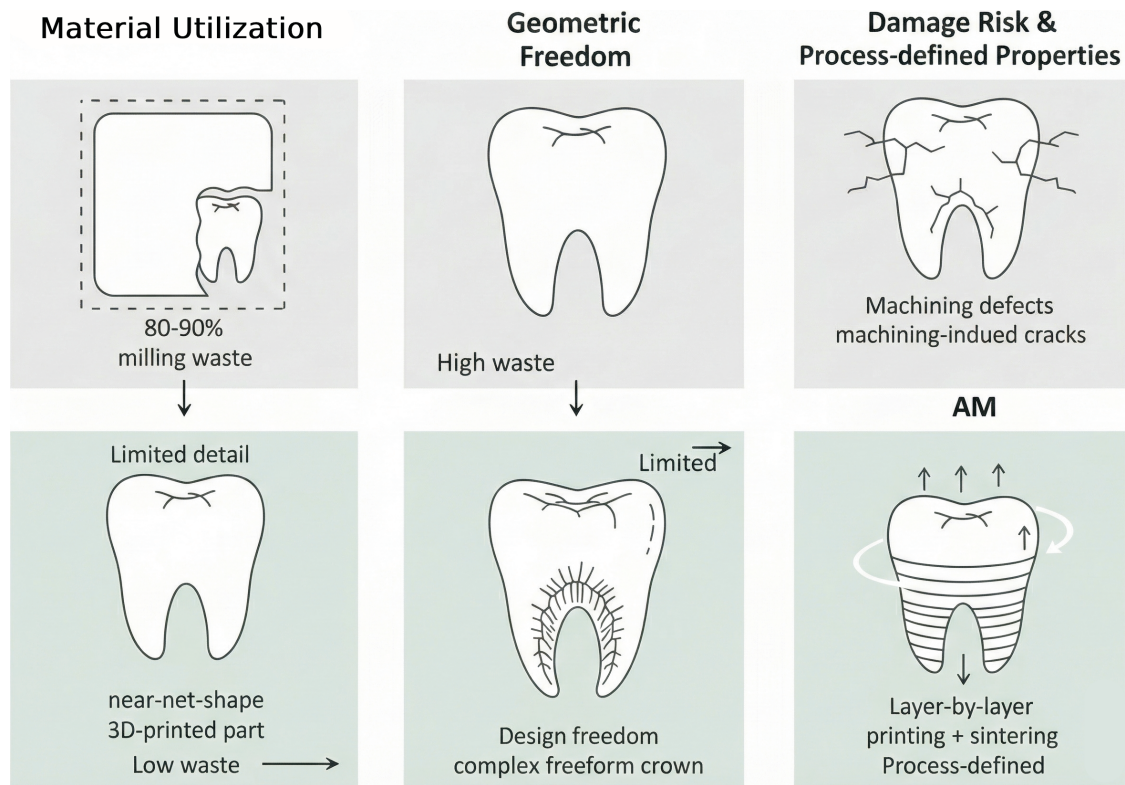


Fig. 1: Value proposition matrix: SM vs AM zirconia (waste, geometry, damage risk; process-defined properties).

The success of any VPP process is critically dependent on the formulation and stability of the ceramic slurry. This is a complex colloidal system where multiple components must be balanced to ensure both printability and the quality of the final sintered part. The primary constituent is the sub-micron zirconia powder, where particle size, distribution, and morphology must be carefully controlled to achieve a high solid loading (typically > 50 vol%) while maintaining a low, processable viscosity¹¹. In practice, SLA/DLP zirconia slurries formulated for reliable layer recoating typically exhibit apparent viscosities below about 5 Pa·s at shear rates of 30–50 s⁻¹, with optimized systems often reported in the 1–3 Pa·s range at ~30 s⁻¹ for solid loadings of around 40–45 vol%. The liquid phase is a photosensitive resin system comprising monomers, oligomers, and photoinitiators that polymerize upon exposure to a specific wavelength of light. A significant challenge in VPP of ceramics is light scattering; the opaque zirconia particles scatter the UV light, which can limit the curing depth, reduce resolution, and lead to poorly defined features or weak interlayer bonding¹⁴. To counteract particle agglomeration and gravitational settling, which would destroy the homogeneity of the slurry, various additives such as dispersants and stabilizers are incorporated. These additives adsorb onto the particle surfaces, creating repulsive forces that keep the particles evenly suspended in the resin¹⁵.

Despite their high precision, VPP-fabricated parts are susceptible to characteristic defects that are a major focus of ongoing research. The layer-by-layer nature of the process can result in a “stair-step” effect on curved or angled surfaces, which increases surface roughness. More

critically, insufficient light penetration or suboptimal slurry formulation can lead to poor adhesion between layers, creating planes of weakness and introducing inter-layer porosity. After sintering, this can manifest as delamination or act as an origin for fracture under mechanical load¹⁶. The removal of the polymer binder during the subsequent debinding stage must also be managed carefully to prevent the rapid formation of gases that can cause cracks or bloating in the part.

(2) Material Extrusion (MEX) technologies (FDM and Robocasting):

Material extrusion represents a fundamentally different approach to AM, relying on the controlled deposition of a viscous material through a nozzle to build a part layer-by-layer. The two main variants relevant to zirconia fabrication are Fused Deposition Modeling (FDM) and Robocasting.

Fused deposition modeling (FDM), also known as fused filament fabrication (FFF), is a widely accessible technology that utilizes a thermoplastic filament as its feedstock. For ceramic applications, this filament is a composite material, consisting of fine zirconia powder (typically 45–65 vol%) homogeneously dispersed within a polymer binder system, such as polylactic acid (PLA) or ethylene-vinyl acetate (EVA)¹⁷. The process involves feeding the filament into a heated extruder, where the binder is softened to a molten state and then extruded through a fine nozzle to deposit a “road” of material. The part is built up by depositing these roads in a predetermined pattern for each layer¹⁸. A critical challenge in FDM of ceramics is achieving high solid loading in the filament; while a higher ceramic content is desirable for achieving a dense final part with minimal shrinkage,

it also makes the filament more brittle and difficult to print without fracturing¹⁹. The most significant hurdle for FDM, however, is ensuring adequate bonding between adjacent layers and roads. The thermal nature of the process can lead to the formation of voids at these interfaces, creating significant, oriented porosity that results in anisotropic mechanical properties. The strength in the vertical (Z) build direction is often substantially lower than in the horizontal (X-Y) plane, which is a major concern for load-bearing dental restorations²⁰.

Robocasting, or direct ink writing (DIW), is a more direct extrusion method that deposits a highly concentrated, paste-like ceramic “ink” at room temperature²¹. The success of this technique hinges on the rheological properties of the ink, which must exhibit shear-thinning behavior. This means the ink behaves like a fluid when under pressure as it passes through the nozzle, allowing it to be extruded smoothly, but immediately becomes a self-supporting solid upon deposition, retaining its shape without slumping²². These inks are typically aqueous-based and can achieve very high solid loadings of zirconia powder (up to 90 wt%) through the use of gelling agents, like Pluronic F-127, and dispersants. A key advantage of robocasting is the potential to formulate binder-free inks, which greatly simplifies the subsequent debinding step and reduces the risk of defects associated with binder burnout²³. However, the high water content in these pastes introduces challenges related to drying; non-uniform drying can induce significant internal stresses, leading to warping and cracking, particularly in larger or more complex geometries²⁴.

(3) Binder Jetting (BJ) technology

Binder jetting (BJ) operates on a principle distinct from VPP and MEX. Instead of using a feedstock that contains both ceramic and binder, BJ separates these two components. The process begins with a roller spreading a thin, uniform layer of dry zirconia powder across a build platform. An inkjet-style print head then moves across the powder bed, selectively depositing droplets of a liquid binder onto the powder, “gluing” the particles together to form the cross-section of the part²⁵. The platform is then lowered, a new layer of powder is spread, and the process repeats. The unbound powder in the bed acts as a natural support structure for overhanging features²⁶. The binder systems used can vary, with common examples including acrylic-based binders or aqueous solutions containing water-soluble adhesives like polyvinyl alcohol (PVOH)²⁷.

The primary characteristic of the BJ process is that the as-printed “green” part is extremely fragile, consisting of ceramic particles held together by a minimal amount of binder, and possesses very high porosity²⁸. Consequently, the properties of the final component are overwhelmingly determined by the post-processing steps. After printing, the green part must be carefully excavated from the powder bed and cured, typically at a low temperature, to gain sufficient handling strength. This is followed by a lengthy and complex thermal cycle for debinding and sintering to achieve densification²⁹. In some appli-

cations, infiltration of the porous sintered part with a secondary material, such as glass, is used to fill voids and improve strength, though this fundamentally alters the material composition and is generally not suitable for high-strength monolithic zirconia restorations³⁰. While BJ offers potential for high throughput and scalability, making it attractive for industrial production, achieving the near-full density (>99 %) and fine surface finish required for reliable dental restorations remains a significant challenge³¹.

(4) Debinding and sintering

It is imperative to recognize that regardless of the AM technology used the initially printed object is not a ceramic but a composite “green” part. This part consists of zirconia particles held in a matrix of a polymer binder. To transform this green part into a dense, functional ceramic restoration, it must undergo a critical two-stage thermal post-processing treatment: debinding and sintering. The success of the entire AM workflow is often dictated by the precision with which these steps are executed.

Debinding is the process of thermally removing the polymer binder from the green part. This must be done with extreme care, using slow heating rates in a controlled atmosphere. If heated too quickly, the polymer decomposes into gaseous products that can build up internal pressure, causing defects such as bloating, cracks, or even catastrophic failure of the part³². Historically, this step has been a major bottleneck, with debinding cycles for complex parts taking anywhere from 20 to over 100 hours, a time frame wholly incompatible with chair-side or even next-day dentistry. This has spurred significant innovation, including ultrafast thermal debinding strategies for VPP-fabricated zirconia that use vacuum pyrolysis and porous graphite felt setters to achieve complete binder removal in less than 30 minutes for small, thin-walled test parts with millimeter-scale section thickness, while maintaining microstructural integrity. For bulky crowns, multi-unit FDPs or connectors with larger cross-sections, however, the longer diffusion paths and higher risk of gas-pressure build-up still necessitate more conservative multi-hour debinding schedules to avoid cracking or bloating, so current <30-minute protocols should be regarded as size-dependent, experimental workflows rather than universally applicable clinical cycles.

Following debinding, the now-porous “brown” part undergoes sintering. During this stage, the part is heated to a high temperature (typically 1 450–1 550 °C), just below the melting point of zirconia. At this temperature, atomic diffusion occurs across the boundaries of the zirconia particles, causing them to fuse together, eliminating porosity and densifying the structure³³. This process is accompanied by significant and non-uniform volumetric shrinkage, often in the range of 20–25 %, which must be precisely predicted and compensated for in the initial digital design to achieve an accurately fitting final restoration³⁴. The sintering profile – including heating rates, final temperature, and dwell time – is a critical control parameter, as it directly influences the

final density, grain size, and crystalline phase composition of the zirconia, which in turn dictate its mechanical and aesthetic properties³⁵. The complexity and sensitivity of these post-processing steps underscore a key reality of ceramic AM: the overall quality of the restoration is determined not by a single step, but by the integrity of the entire multi-stage workflow. A flaw introduced at any point, from slurry preparation to final sintering, can compromise the final product, making a holistic, system-level approach to process control and validation essential for clinical success³⁶. A concise comparison of these additive manufacturing technologies is presented in **Table 1**, summarizing their underlying principles, feedstock types, advantages, challenges, and typical resolutions. This overview highlights the trade-offs between precision, scalability, and material versatility across different AM routes³⁷. Notably, while VPP methods excel in producing fine, high-resolution features ideal for single-unit restorations, extrusion-based and binder jetting processes offer greater throughput but at the

cost of surface finish and mechanical uniformity³⁸. Such distinctions are critical when selecting the most suitable fabrication route for specific clinical and laboratory applications. Given debinding's bottleneck and sintering's shrinkage sensitivity, **Fig. 2** maps a temperature-time "safe window" with coupled failure modes (foaming, cracking, warpage) and control levers (heating rates, atmosphere, part venting, green density). An inset compares conventional multi-hour debinding to accelerated pathways (<30 min) and highlights trade-offs in defect risk versus throughput³⁹. Overall, multi-step low-rate debinding (typically 20–50 h) widens the safe window and favors dimensional stability, whereas accelerated or debinding-lean routes (<30 min) markedly increase throughput but require tighter control of green density, part geometry and gas venting to avoid defects, as emphasized in recent debinding reviews and process-window analyses. This process window underpins later discussions of density, grain size, and phase composition.

Table 1: Comparative overview of key additive manufacturing technologies for zirconia fabrication, highlighting their working principles, feedstock characteristics, advantages, processing challenges, and typical printing resolutions.

Tech-nology	Principle	Feedstock Material	Key Advantages	Primary Challenges	Typical Resolution	Reference
SLA	Vat photopolymerization using a laser to cure a liquid slurry	Zirconia powder suspended in a photosensitive liquid resin (slurry)	High accuracy and resolution; excellent surface finish	Slow build speed; slurry stability; light scattering by ceramic particles	25–100 μm	⁴⁰
DLP	Vat photopolymerization using a digital projector to cure an entire layer at once	Zirconia powder suspended in a photosensitive liquid resin (slurry)	High build speed; good accuracy and surface finish	Slurry stability; light scattering; potential for layer-to-layer distortion	30–100 μm	⁴¹
FDM	Material extrusion of a thermoplastic filament containing ceramic powder	Composite filament of zirconia powder and a polymer binder (e.g., PLA, EVA)	Low-cost equipment; wide material availability; simple operation	Poor interlayer bonding leading to anisotropy and porosity; lower resolution	100–300 μm	⁴²
DIW	Material extrusion of a highly concentrated ceramic paste at room temperature	High-viscosity "ink" of zirconia powder in an aqueous gel	Potential for binder-free processing; high solid loading; material versatility	Drying-induced stresses causing warping and cracking; lower resolution	100–400 μm	⁴³
BJ	Inkjet printing of a liquid binder onto a bed of ceramic powder	Dry zirconia powder and a separate liquid binder	High throughput/scalability; no support structures needed; material versatility	Low green part strength; high porosity requiring extensive post-processing; poor surface finish	80–200 μm	⁴⁴

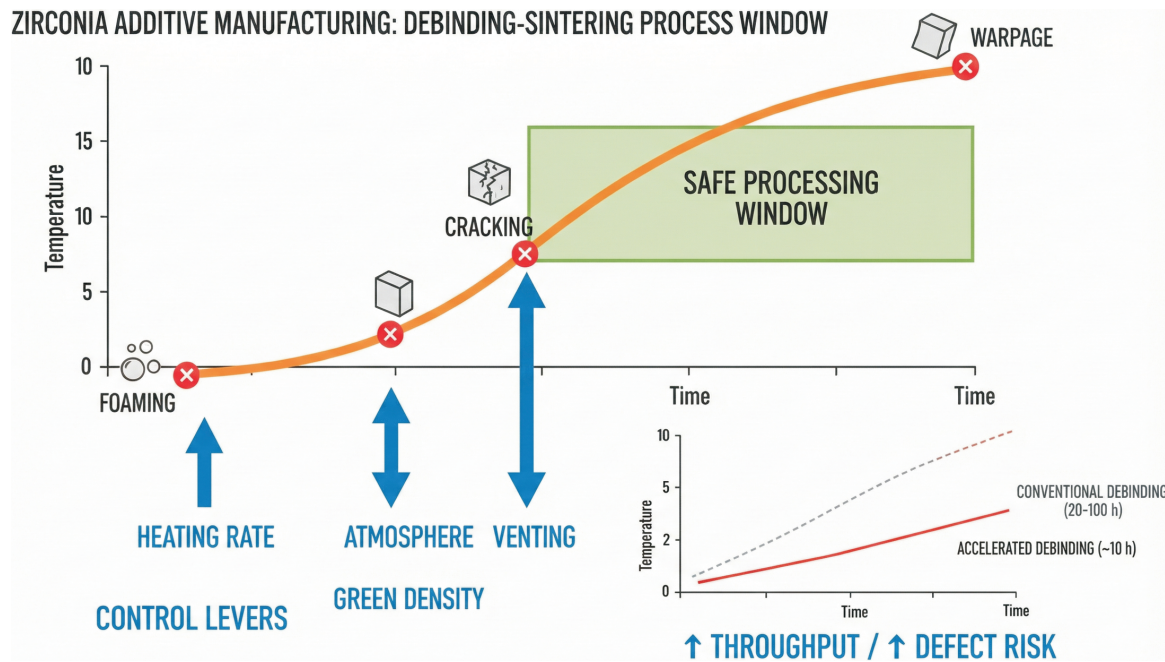


Fig. 2: Debinding-sintering process window with failure modes and mitigation levers.

III. Material Properties and Microstructure: AM Zirconia versus the SM Gold Standard

The central question governing the clinical viability of additively manufactured zirconia is whether it can consistently match or exceed the well-characterized and clinically proven properties of its subtractively manufactured counterpart. The literature presents a complex and often contradictory picture, where the performance of AM zirconia is highly dependent on the specific technology, material formulation, and processing parameters employed. This section critically evaluates the existing evidence on mechanical performance, microstructural characteristics, dimensional accuracy, and aesthetic properties, directly addressing the contentious debate over AM zirconia's readiness to challenge the SM gold standard.

(1) Mechanical performance and reliability: a contentious debate

The mechanical integrity of a dental restoration is paramount for its long-term clinical survival. The debate surrounding AM zirconia is most intense in this domain, with conflicting reports on its fundamental mechanical properties.

The flexural strength, a key indicator of a material's ability to resist fracture under bending loads, is a primary point of controversy. Several studies have reported encouraging results, suggesting that AM zirconia, particularly when fabricated via advanced VPP techniques like SLA, can achieve biaxial flexural strength values comparable to or even exceeding those of milled 3Y-TZP, comfortably satisfying the criteria for Class 5 dental ceramics (>800 MPa) according to the ISO 6872 standard¹⁴. For instance, one study found no significant difference in flexural strength between SLA-fabricated and SM-fabricated zirconia specimens⁴⁵. However, this optimistic view is challenged by a substantial body of

evidence, including systematic reviews and meta-analyses, which concludes that AM zirconia generally exhibits significantly lower flexural strength than SM zirconia⁴⁶. One direct comparison found that CNC-milled zirconia had a flexural strength of 1140 MPa, whereas VPP-printed specimens (SLA and DLP) reached only 894 MPa and 831 MPa, respectively⁴⁷. This discrepancy in the literature highlights the profound influence of the entire manufacturing workflow on the final properties; variations in slurry composition, printing parameters, and especially the debinding and sintering cycles can lead to vastly different outcomes even when using the same base material.

In contrast, the data on fracture toughness and hardness show greater consensus. Multiple independent studies have demonstrated that when AM zirconia is properly processed to achieve high density, its fracture toughness and Vickers hardness are comparable to those of SM zirconia⁴⁸. This suggests that the intrinsic material properties can be replicated. However, the reliability and durability under clinical conditions are more complex. Dynamic fatigue testing, which simulates the repetitive, low-level forces of mastication, provides a more clinically relevant assessment than static strength tests. Here, the evidence points to a clear deficiency in current AM materials. Studies have shown that AM zirconia has a lower fatigue failure strength and withstands significantly fewer loading cycles before failure compared to SM zirconia⁴⁹. This inferior fatigue performance is directly attributed to the higher prevalence of inherent microstructural defects in AM parts, such as porosity and weak interlayer interfaces, which act as stress concentrators and initiation sites for fatigue cracks. **Fig. 3** links process defects to reliability through a "Process-Defect-Property" causal map: (i) interlayer weakness, entrapped porosity, and incomplete burnout act as critical flaws; (ii) these amplify under cyclic mastication, lowering fatigue strength;

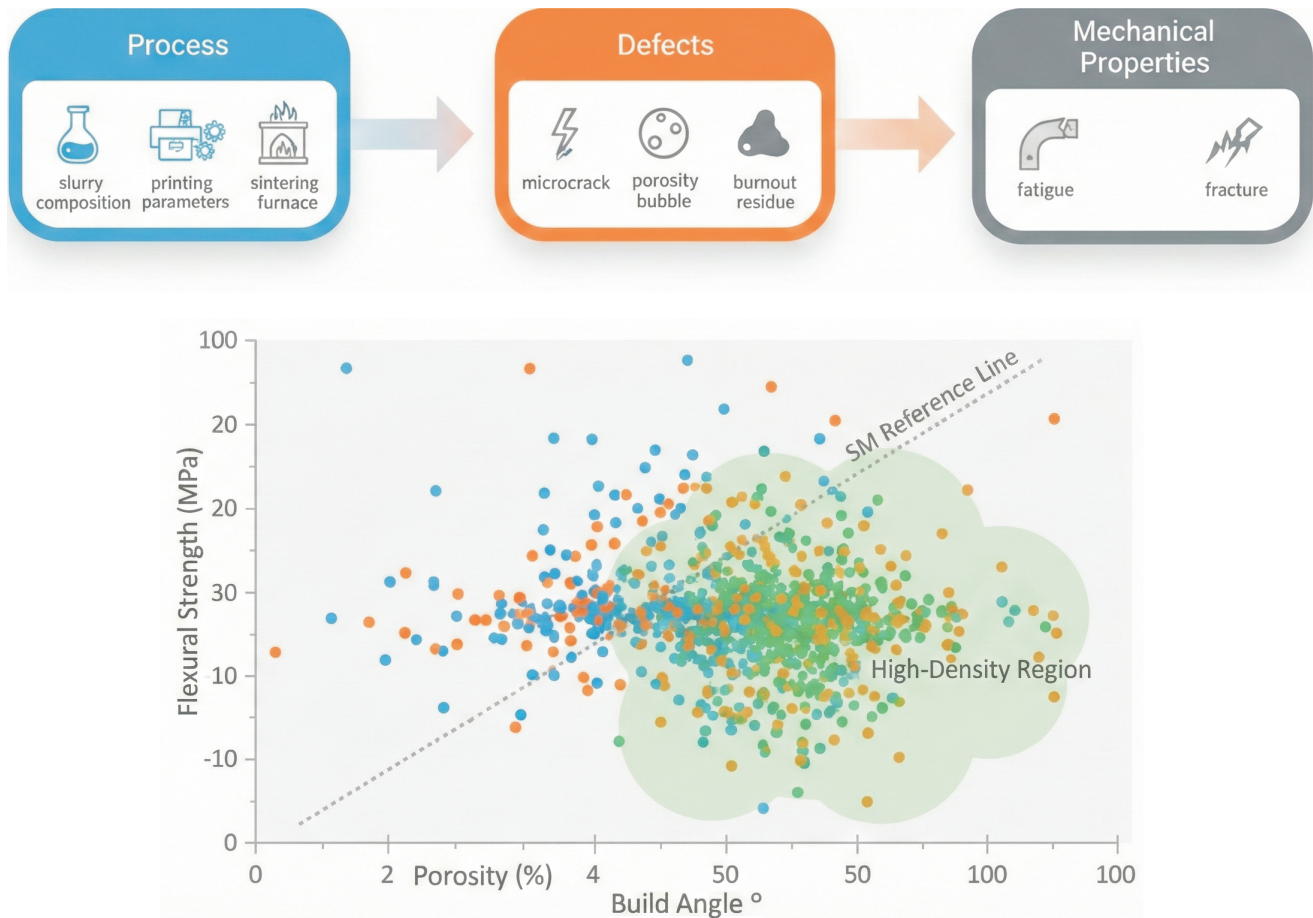


Fig. 3: Process-defect-property map and strength domains vs porosity/orientation.

(iii) orientation-dependent roads/layers reduce the Weibull modulus. A companion plot visualizes flexural strength versus porosity and build angle, delineating regions where AM approaches SM performance when density is $\geq 99\%$ and interlayer bonding is optimized.

A fundamental disadvantage of most AM processes is the introduction of structural anisotropy. The mechanical properties of a printed part can vary significantly depending on its orientation during the build process. It is well-documented that parts printed with their longest axis parallel to the build plate (0° or horizontal orientation) tend to exhibit higher flexural strength than those printed perpendicular to it (90° or vertical orientation)⁵⁰. This is because in the horizontal orientation, bending forces are applied along the strong, continuous printed layers, whereas in the vertical orientation, forces are applied across the weaker interlayer junctions. This anisotropy complicates the design and fabrication process and introduces a level of unpredictability not present in the isotropic, industrially pressed blocks used for milling. The reliability of a material, often quantified by the Weibull modulus, is a measure of the consistency of its strength. The wider scatter in strength data often reported for AM zirconia translates to a lower Weibull modulus, indicating lower reliability and a higher probability of premature failure compared to the highly consistent SM zirconia⁵¹.

(2) Microstructure, porosity, and phase stability

The macroscopic mechanical properties of a ceramic are a direct reflection of its microscopic structure. The primary factor differentiating AM and SM zirconia at the microstructural level is the presence and nature of porosity. While industrially fabricated SM blanks achieve near-full theoretical density ($>99.5\%$), AM parts are prone to various forms of porosity that can severely compromise their mechanical performance⁵². These defects can arise from multiple sources throughout the workflow: air bubbles trapped in the initial slurry, incomplete burnout of the polymer binder leaving behind voids, poor fusion between layers or extruded filaments (particularly in MEX), and insufficient particle consolidation during sintering⁵³. Even small amounts of porosity can act as critical flaws, dramatically reducing the strength and fatigue resistance of the ceramic⁵⁴.

When high density is achieved, the grain structure of AM zirconia can be comparable to, or even finer and more uniform than, that of SM zirconia, which is generally beneficial for mechanical properties⁵⁵. However, another critical aspect of long-term performance is phase stability. 3Y-TZP relies on a metastable tetragonal crystal structure for its toughness. In the humid environment of the oral cavity, this tetragonal phase can slowly transform into the monoclinic phase, a phenomenon known as low-temperature degradation (LTD) or aging. This transformation is

accompanied by a volumetric expansion that creates intergranular microcracks, leading to surface roughening and a progressive loss of strength⁵⁴. Beyond grain-boundary microcracking, the residual inter- and intralayer porosity that is more frequently observed in AM zirconia than in dense SM blanks can provide additional pathways for water penetration, thereby accelerating the tetragonal-to-monoclinic transformation within the bulk. Experimental studies on Y-TZP with intentionally porous surface layers have shown markedly faster LTD kinetics in the porous regions compared with the dense core when exposed to water, underscoring the role of connected porosity in promoting aging. After 5 h at 134 °C, the monoclinic fraction increased from 0 % to 19.6 % in SLA and from 0 % to 9.6 % in DLP zirconia, compared with 0 % to 2.3 % in the milled control group, indicating faster t→m transformation kinetics in some VPP materials under identical aging conditions. There is emerging evidence that some AM zirconia formulations may be more susceptible to LTD than their SM counterparts. Studies have reported that certain DLP-fabricated zirconia materials exhibit a higher initial cubic phase content and a significantly faster rate of tetragonal-to-monoclinic phase transformation upon artificial aging⁵⁶. This raises serious concerns about the long-term stability and clinical longevity of these materials, necessitating further rigorous investigation.

(3) Accuracy and marginal fit of restorations

For any dental crown or FDP, a precise fit is a non-negotiable clinical requirement. Poor marginal adaptation leads to cement dissolution, microleakage, secondary

caries, and periodontal inflammation. A marginal gap of 120 µm is often cited as the maximum clinically acceptable limit. This is an area where SM technology currently holds a distinct advantage. A consistent finding across numerous *in-vitro* studies is that 3D-printed zirconia crowns exhibit statistically significant larger marginal and internal gaps compared to milled crowns⁵⁷. For example, one study using a high-precision triple-scan protocol measured mean marginal gaps of 87.7 ± 7.4 µm for 3D-printed crowns versus 57.5 ± 7.0 µm for milled crowns and internal gaps of 107.4 ± 4.9 µm versus 86.6 ± 7.6 µm ($n = 12$ per group, $p < 0.001$), all below the 120 µm clinical threshold⁵⁸. Accuracy is a chain, not a node. **Fig. 4** decomposes the error budget from voxel/pixel cure, staircase geometry, polymerization and thermal shrinkage, to scanner-CAD-CAM registration. It overlays compensation strategies (XY offset, global scale factors, orientation-specific scaling) and shows their cumulative effect on predicted marginal/internal gaps. In practice, orientation-specific scaling applies different pre-sintering enlargement factors along the X, Y, and Z axes, derived from experimentally measured, orientation-dependent shrinkage of the printed zirconia. This allows crowns printed at different build orientations to converge towards the same target dimensions after sintering, as demonstrated in recent ceramic SLA and DLP zirconia studies that use axis-specific shrinkage factors to keep linear deviations very close to the nominal design. The pipeline explains how AM crowns can remain clinically acceptable while still trailing SM in mean fit.

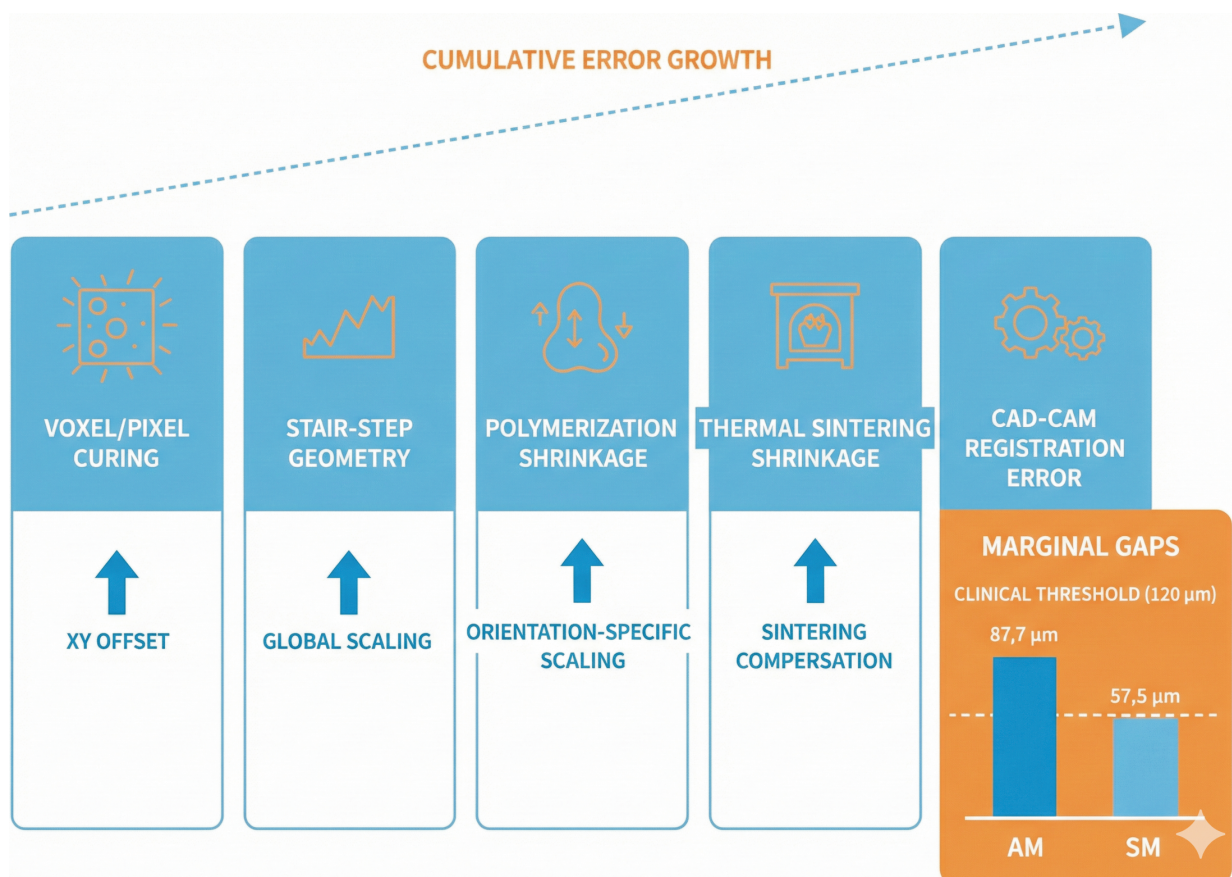


Fig. 4: Dimensional accuracy pipeline: error budget and compensation strategies.

The sources of inaccuracy in AM are multifaceted. They include the inherent resolution limits of the printer, shrinkage of the resin during photopolymerization, the “stair-step” effect on non-parallel surfaces, and, most significantly, the large and often anisotropic volumetric shrinkage that occurs during sintering⁵⁹. This shrinkage must be accurately predicted and compensated for by scaling up the initial digital design, a process that is far more complex than for the standardized industrial SM blanks. However, a crucial point of discussion arises from these findings. While statistically inferior, the marginal fit values reported for many AM crowns still fall within the clinically acceptable range of $<120\ \mu\text{m}$ ⁶⁰. This creates a central debate: is a “clinically acceptable” fit sufficient to justify adopting a new technology, especially when the existing technology provides a demonstrably superior fit? The answer may depend on the specific clinical situation and the potential workflow advantages of AM. Researchers are actively working to improve accuracy through software optimization, such as applying XY-offset compensation parameters to counteract process-related deviations⁶¹.

(4) *Aesthetic properties: translucency and color*

The aesthetic success of a ceramic restoration is largely determined by its ability to mimic the translucency and color of natural teeth. Modern zirconia materials used for SM have made great strides in this area, with the development of “high-translucency” (4Y-PSZ) and “ultra-translucency” (5Y-PSZ) variants that contain a higher percentage of the cubic phase, which reduces light scattering⁶². The translucency of any ceramic is highly sensitive to its microstructure; factors such as grain size, grain boundaries, and, most importantly, porosity, all contribute to light scattering and increase opacity. Consequently, any residual porosity remaining after the sintering of an AM part will inevitably compromise its translucency⁶³. Unless otherwise stated, translucency percentages in this section refer to total visible-light transmittance ($T_t\%$) through 1.0-mm-thick zirconia specimens measured with a spectrophotometer under standardized illumination. Currently, commercially available zirconia slurries for VPP typically produce restorations with a lower level of translucency (reported around 30 % total visible-light transmittance) compared to the high-translucency milled blocks they aim to replace.

Currently, commercially available zirconia slurries for VPP typically produce restorations with a lower level of translucency (reported around 30 %) compared to the high-translucency milled blocks they aim to replace⁶⁴. This limits their application in the highly aesthetic anterior region. However, this is an area of active development, with manufacturers reporting the imminent release of

more translucent slurries that achieve ~40 % total visible-light transmittance, bringing them closer to the aesthetic standards of milled ceramics⁶⁵. Furthermore, AM offers unique possibilities for aesthetic innovation that are impossible with SM. For example, FDM has been used to successfully fabricate gradient-colored zirconia discs by co-extruding two different colored filaments, creating a smooth transition from a more opaque, chromatic cervical region to a more translucent incisal edge within a single, monolithic structure⁶⁶. In that feasibility work, a standard FDM printer was modified with a custom dual-feed hot-end and used in combination with commercially available white and black zirconia filaments originally developed for technical applications (e.g. Zetamix), rather than a dedicated dental FDM system, and no regulatory approval for clinical use has yet been established. This capability to create functionally graded aesthetic materials represents a significant potential advantage for AM in the future.

The collective evidence suggests a nuanced conclusion. For applications where maximum mechanical performance, reliability, and precision are the absolute priorities, such as long-span posterior FDPs, the material superiority of SM zirconia remains unchallenged. However, for a significant portion of common dental restorations, like single posterior crowns, the properties of AM zirconia are approaching a level of “workflow sufficiency.” While perhaps not superior in absolute terms, its mechanical and physical properties may be adequate for clinical service, allowing the significant workflow advantages of AM – reduced waste, design freedom, and potential for chair-side fabrication – to become the deciding factors for its adoption. The focus of the debate is thus shifting from a simple comparison of material properties to a more complex evaluation of the overall value proposition of the entire manufacturing workflow for specific clinical indications. A quantitative synthesis of the published literature further underscores these trends. **Table 2** consolidates key comparative studies on flexural strength, fracture toughness, and hardness of AM and SM zirconia, highlighting both the encouraging performance of optimized VPP systems and the persistent fatigue and reliability gap. For each study, we now also report the test standard, specimen geometry and thickness, printing and build orientation (with layer thickness), sintering peak temperature/hold time, sample size and basic statistics to clarify how these methodological factors shape the reported means. Meanwhile, **Table 3** summarizes the dimensional accuracy data, confirming that while AM restorations consistently show larger marginal and internal gaps, they remain clinically acceptable in most cases. These tabulated findings emphasize that the performance gap is narrowing but not yet fully closed, particularly for fatigue resistance and precision fit.

Table 2: Summary of key comparative studies evaluating the mechanical properties of AM versus SM zirconia, including flexural strength, fracture toughness, and hardness, highlighting performance variability across different AM technologies and materials.

Study (Author, Year)	AM Technology	AM Material	SM Material (Control)	Flexural Strength (MPa) (AM vs SM)	Fracture Toughness (MPa·m ^{1/2}) (AM vs SM)	Vickers Hardness (HV) (AM vs SM)	Key Finding/ Conclusion
67	DLP	3D-printed Zirconia	Milled Zirconia	1658 vs 1890 (Static) 1224 vs 1642 (Fatigued)	Not reported	Not reported	SM crowns had significantly higher fracture resistance than AM crowns, both before and after cyclic loading.
47	SLA, DLP	3D-printed Zirconia	CNC Milled Zirconia	894 (SLA), 831 (DLP) vs 1140	No significant difference	Comparable	SM zirconia has superior static flexural strength and dynamic fatigue resistance compared to VPP-printed zirconia.
60	SLA, DLP	Various AM Zirconia	Milled Zirconia	Ranged from 1441.94 to 1532.53 (AM) vs 1473.87 (SM)	Not reported	Ranged from 1441.94 to 1532.53 (AM) vs 1473.87 (SM)	Some AM groups demonstrated higher hardness than the SM group. All values were clinically ideal (> 1200 HV).
36	SLA	3DMix Zirconia	Prettau Zirconia	No significant difference	No significant difference	Not reported	AM and SM zirconia showed similar phase composition and comparable flexural strength and fracture toughness.
68	Meta-Analysis	Various AM Zirconia	Various SM Zirconia	Significantly lower for AM (g = -2.868)	Less favorable for AM	Less favorable for AM	SM had better overall accuracy, higher flexural strength, and more favorable hardness and fracture toughness.
24	SLA	LithaCon 3Y 230, 3D Mix ATZ	LAVA Plus 3Y-TZP	LithaCon: Lower 3D Mix ATZ: Higher	Not Reported	Not Reported	AM ATZ significantly outperformed SM 3Y-TZP. AM 3Y-TZP showed comparable properties to SM zirconia.

Table 3: Marginal and internal fit of zirconia crowns (AM vs SM).

Study (Author, Year)	AM Technology	Measurement Method	Marginal Gap (μm) (AM vs SM)	Internal Gap (μm) (AM vs SM)	Clinically Acceptable? (< 120 μm)
2	Meta-Analysis (mostly DLP)	Various	Higher for AM (g = 1.05)	Higher for AM	Yes, but less accurate than SM.
58	VPP (Lithoz)	Triple Scan Protocol	87.7 vs 57.5	107.4 vs 86.6	Yes, both were within limits.
69	DLP	VMGT & SRT	80 vs 60 (VMGT) 100 vs 60 (SRT)	Significant difference except for Axial Gap	Yes, both were acceptable.
60	SLA, DLP	Replica Technique	42.83 – 81.95 vs 48.45	Significant differences between groups	Yes, all measurements were acceptable.
61	DLP	Triple Scan Protocol	97.0 (no offset) 77.6 (with offset)	121.2 (no offset) 90.4 (with offset)	Yes, fit improved significantly with software offset correction.

IV. Clinical Applications, Performance, and Biocompatibility

Translating the material science of additively manufactured zirconia into clinical practice requires a rigorous evaluation of its performance in simulated oral environments, a thorough assessment of its biological safety, and ultimately, validation through clinical trials. While the technology is rapidly advancing, its readiness for widespread patient care is contingent upon demonstrating not only adequate mechanical and aesthetic properties but also long-term durability and biocompatibility. This section assesses the current state of AM zirconia from a clinical perspective, examining its prospective applications, *in-vitro* performance, biological safety profile, and the critical gap in *in-vivo* evidence.

(1) Current and prospective clinical applications

The potential applications for AM zirconia in dentistry are broad, mirroring and in some cases expanding upon those of milled zirconia. The most immediate and widely researched application is the fabrication of single-unit monolithic crowns and veneers. As the technology matures and demonstrates greater reliability, its use is expected to extend to multi-unit FDPs and full-arch implant-supported frameworks. However, the true transformative potential of AM lies in its ability to create geometries that are difficult or impossible to produce with subtractive methods. This includes patient-specific dental implants with complex surface topographies designed to enhance osseointegration without the need for potentially damaging post-fabrication treatments like sandblasting. Furthermore, the geometric freedom of AM allows for the design of implant abutments and frameworks with highly optimized internal structures for improved stress distribution and reduced weight, pushing the boundaries of prosthetic design.

(2) *In-vitro* performance and tribology

While standardized tests on simple geometric specimens provide essential baseline data, *in-vitro* studies on anatom-

ically shaped restorations offer more clinically relevant insights into performance. When full-contour crowns are subjected to compressive load until fracture, a consistent trend emerges: milled zirconia crowns exhibit significantly higher fracture resistance than their 3D-printed counterparts⁷⁰. This finding aligns with the lower flexural strength and higher defect population observed in AM materials.

To better simulate the oral environment, researchers employ cyclic loading to induce fatigue. These studies are critical, as most clinical failures are due to fatigue rather than a single overload event. The evidence clearly shows that cyclic loading significantly degrades the fracture resistance of both AM and SM crowns. However, the degradation is often more pronounced for AM restorations, further highlighting the role of manufacturing-induced defects as stress concentrators that accelerate fatigue crack propagation⁶⁷. Despite this reduction in strength, Afrashtehfar *et al.*⁷¹ reported that DLP-printed crowns still fractured at $1\,658 \pm 247$ N under static loading and $1\,224 \pm 177$ N after 1.2 million fatigue cycles, compared with $1\,890 \pm 257$ N and $1\,642 \pm 196$ N for milled crowns ($n = 9$ per group, $p < 0.001$)⁶⁷; given that maximum posterior masticatory forces are typically <900 N, these values indicate that current AM crowns can still exceed functional demands for single-unit posterior restorations.

The tribological behavior – specifically, the wear of opposing enamel – is a critical factor for the long-term success of any restorative material. While research on the wear characteristics of AM zirconia is still nascent, valuable inferences can be drawn from the extensive literature on SM zirconia. It is well-established that the surface finish of zirconia is the dominant factor in its wear behavior. Numerous *in-vitro* studies have conclusively shown that a well-polished monolithic zirconia surface is exceptionally kind to opposing enamel, often causing less wear than enamel-on-enamel contact⁷². In stark contrast, glazed zirconia is highly abrasive. The thin glaze layer wears away quickly in function, exposing the rougher, unpolished ceramic underneath,

which then acts as an aggressive abrasive against the opposing dentition⁷³. This has profound implications for AM zirconia. The as-sintered surface of an AM part, with its characteristic layer lines, is likely to be abrasive. Therefore, achieving a final, meticulously polished occlusal surface will be a non-negotiable step in the clinical workflow to prevent iatrogenic damage to the opposing dentition. The comparative mechanical performance of AM and SM zirconia crowns under static and cyclic loading conditions is summarized in **Table 4**. The data clearly highlight the superior fracture strength of milled crowns, especially after fatigue loading, emphasizing the critical influence of microstructural defects on the fatigue behavior of AM restorations. Notably, however, the average fracture loads of AM crowns remain above physiological masticatory forces, supporting their clinical potential for single-unit applications. Bridging bench to chairside, **Fig. 5** proposes a translation roadmap: indication tiers (single crowns → multi-unit frameworks), required evidence (fatigue/aging → RCTs), and a quality system overlay (incoming feedstock control, validated cleaning/UV cure, ISO 10993 cytotoxicity, traceability, ISO 13485 responsibilities). A performance envelope

compares static/fatigue fracture loads to physiological bite forces and annotates the mandatory polished-occlusal finish to minimize antagonistic enamel wear.

(3) Biocompatibility and cytotoxicity

The biological safety of any long-term restorative material is a primary concern. The assessment of AM zirconia's biocompatibility must be considered as a two-stage problem, addressing both the final sintered material and the potential for residues from the manufacturing process.

Zirconia as a bulk material is renowned for its excellent biocompatibility and chemical inertness⁷⁸. *In-vitro* studies on final, fully sintered AM zirconia parts confirm this. They consistently demonstrate high levels of biocompatibility, showing good cell adhesion and proliferation (e.g., with human gingival fibroblasts or osteoblasts) and eliciting minimal inflammatory or cytotoxic responses⁷⁹. These findings suggest that the final ceramic product, once properly processed, is as safe for intraoral use as its milled counterpart.

However, a significant and often overlooked risk lies in the “green” state and the intermediate processing steps.

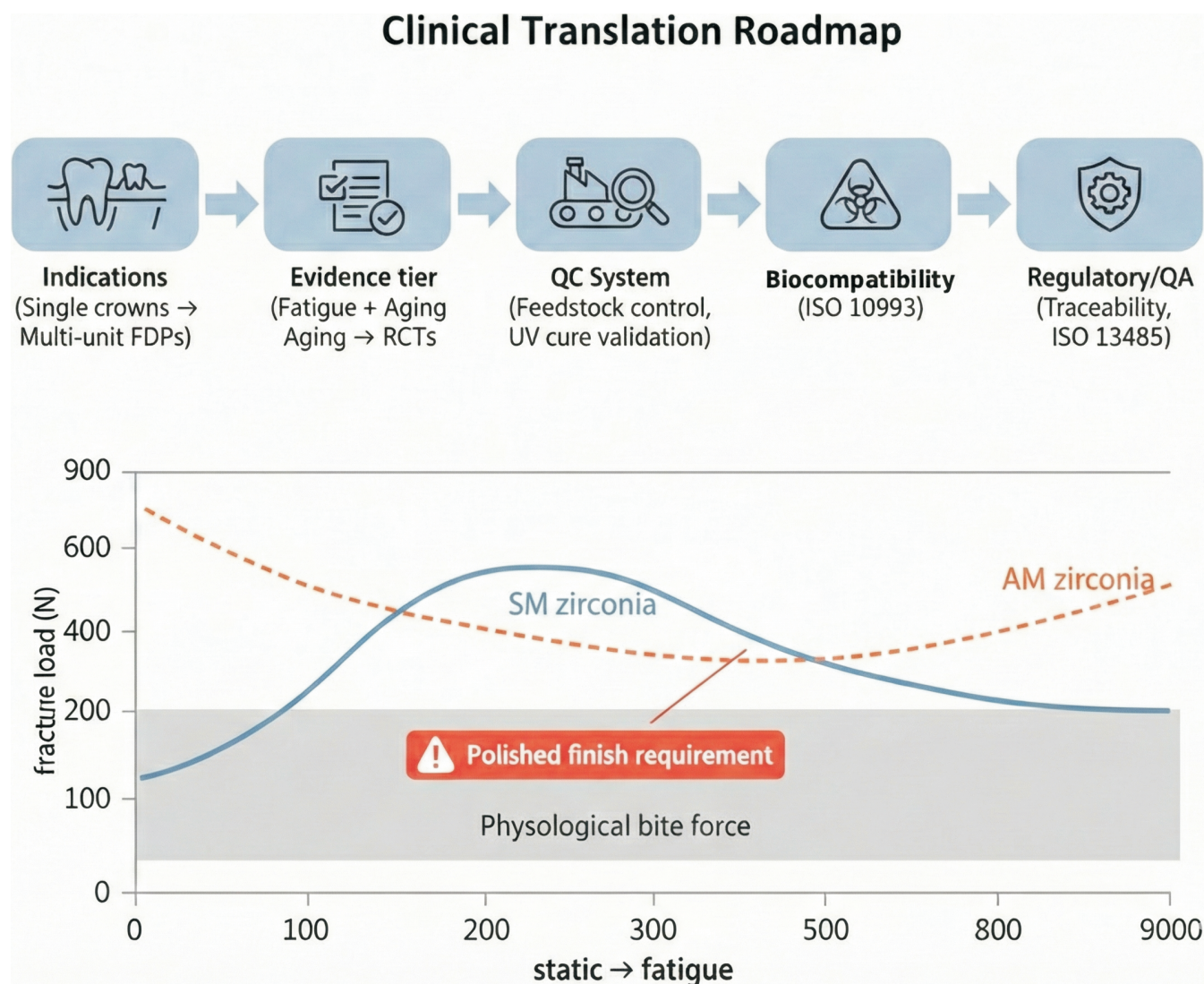


Fig. 5: Clinical translation roadmap with QC/biocompatibility checkpoints and performance envelope.

Table 4: Summary of *in-vitro* mechanical performance of AM and SM zirconia restorations under static and cyclic loading conditions, highlighting fracture load values and key findings relevant to clinical strength and fatigue behavior.

Study (Author, Year)	Fabrication Method (AM vs SM)	Test Condition	Mean Fracture Load (N)	Key Finding
67	DLP vs Milled	Static Load	1658 vs 1890	Milled crowns were significantly stronger.
67	DLP vs Milled	Cyclic loading (1.2 M cycles)	1224 vs 1642	Cyclic loading reduced strength for both, but milled crowns remained significantly stronger.
74	Review	Static load	Milled group demonstrated significantly higher fracture resistance.	Aging reduced strength in both groups.
75	Milled only (HT vs FX vs Prefab)	Static load	HT: 3087, Prefab: 2633, FX: 2483	High-translucency milled zirconia showed the highest fracture resistance. All values exceeded maximum bite forces.
76	AM monolithic vs AM bi-layered	Static load	No significant difference	Both AM crown types showed similar fracture resistance.
77	3D-printed resin vs milled hybrid ceramic	Static load	Fracture strength decreased with thickness for AM, but increased for SM.	All materials had fracture strengths higher than average occlusal forces at all thicknesses.

The photopolymer resins used in VPP and the binders used in MEX and BJ technologies contain a variety of chemical components, including unreacted monomers, oligomers, photoinitiators, and solvents, many of which are known to be cytotoxic⁸⁰. If these components are not completely removed from the green part before sintering, or if they are not fully burned out during debinding, they could potentially remain as trace residues in the final restoration. The leaching of such substances into the oral environment could pose a biological risk. This makes the post-printing cleaning and curing protocols absolutely critical. International standards, such as ISO 10993, provide a framework for the biological evaluation of medical devices and must be applied to the entire AM workflow⁸¹. Encouragingly, studies on photopolymer resins have shown that validated post-processing protocols, such as a double wash in isopropanol (IPA) combined with a final UV cure, can be highly effective at removing leachable cytotoxic residues and ensuring the biocompatibility of the final part⁸². The critical question for clinical translation is whether these essential cleaning steps can be standardized and reliably implemented in a busy dental laboratory or clinical setting to guarantee patient safety for every restoration produced.

The adoption of AM for zirconia represents a significant shift in the manufacturing model for dental restorations. With SM, dental laboratories work with pre-sintered zirconia blocks that are manufactured under stringent industrial quality controls and are typically cleared by regulatory bodies like the FDA or carry a CE mark as medical devices. The lab's primary function is to machine this pre-approved material. With the advent of in-office or in-lab AM, the dental practice or laboratory effectively becomes the manufacturer of

the medical device. They assume responsibility for the entire process chain, from managing the raw material (slurry or filament) and controlling the printing process to executing the critical thermal post-processing and ensuring the final part is properly cleaned and biocompatible. This paradigm shift transfers the burden of quality control and regulatory compliance from the industrial supplier to the end-user. For AM to be adopted safely and responsibly, it will require dental laboratories and clinics to implement robust quality management systems, akin to those required for medical device manufacturing (e.g., ISO 13485). This represents a significant logistical, educational, and regulatory hurdle that may temper the pace of adoption, even as the underlying technology continues to improve. The biological response to AM zirconia and related materials, including both ceramic and photopolymer-based systems, is summarized in Table 5. These results demonstrate excellent *in-vitro* biocompatibility of sintered AM zirconia, while underscoring the cytotoxic potential of unremoved resin residues. This duality reinforces the critical need for standardized post-processing protocols and regulatory oversight in clinical implementation.

(4) Clinical evidence and future outlook

Despite the flurry of *in-vitro* research and rapid technological advancements, the single greatest barrier to the widespread clinical adoption of additively manufactured zirconia is the profound lack of long-term, *in-vivo* clinical evidence. The current body of literature is overwhelmingly dominated by laboratory studies, with a near-complete absence of randomized controlled clinical trials (RCTs) comparing the long-term performance of AM zirconia restorations to the SM gold standard. This

Table 5: Summary of *in-vitro* biocompatibility and cytotoxicity studies on AM zirconia and related materials.

Study (Author, Year)	Material/Process	Cell Line Tested	Assay	Outcome	Compliance with ISO 10993–5
83	ZrO ₂ -ZnO (AM)	<i>In-vivo</i> rabbit model	Histological evaluation (HE staining)	Mild inflammatory response, good biocompatibility with bone tissue.	N/A (<i>In-vivo</i> study)
83	ZrO ₂ /SiO ₂ (AM)	MG63	Cell proliferation	50 wt% zirconia group showed the best proliferation.	Not explicitly stated
84	3D-printed zirconia vs milled zirconia	Human gingival keratinocytes (IHGKs)	Adhesion, cell viability (MTS)	Good biocompatibility for both. AM showed lower cell viability after 96 h of direct contact.	Not explicitly stated, but implies good biocompatibility.
85	3D-printed 3Y-SZ	Human bone-marrow MSCs	Biocompatibility tests	Confirmed high biocompatibility of 3Y-SZ.	Not explicitly stated
80	3D-printed resin (not zirconia)	Human gingival fibroblasts (HGFs)	MTT, LDH, PrestoBlue	Slight cytotoxic effect, but decrease in metabolic activity was < 30 %.	Considered nontoxic according to the norm.
81	3D-printed resins (not zirconia)	Human mesenchymal stem cells (MSCs)	Cell viability	All tested photopolymers reduced cell viability, even certified biocompatible ones.	Certified resins showed minor toxic effects, but none were fully recommended.
82	Photopolymer resins (not zirconia)	Myoblasts, fibroblasts	Cell viability	Post-print cure with double IPA wash effectively reduces cytotoxicity, maintaining ~ 100 % cell viability.	N/A (Process study)

evidence gap makes it difficult for clinicians to confidently adopt the technology for permanent, load-bearing restorations.

In summary, additive manufacturing of zirconia is a technology of immense promise that has matured at an astonishing pace. It has demonstrated the capability to produce dental restorations with mechanical properties and marginal fit that, in many cases, meet the minimum requirements for clinical acceptability. However, the technology as a whole has not yet consistently demonstrated the high level of mechanical reliability, microstructural quality, dimensional precision, and long-term stability that characterizes the established SM workflow. The variability in outcomes reported across the literature underscores that the quality of an AM restoration is a function of an entire, complex workflow, not just the printer itself.

The path forward for AM zirconia requires a concerted effort from researchers, manufacturers, and clinicians. The critical next steps for the field must include: (1) the standardization of feedstock materials and end-to-end processing workflows to reduce variability and ensure consistent outcomes; (2) the development and implementation of non-destructive evaluation (NDE) techniques for quality control of printed and sintered parts, a crucial step for a process with many potential failure points;

(3) comprehensive, long-term *in-vitro* studies focusing on fatigue and hydrothermal aging to better predict clinical longevity; and (4) most importantly, the initiation of well-designed, multi-center, long-term RCTs to validate the promising *in-vitro* findings and establish clear, evidence-based clinical indications for the use of this transformative technology. Only through such rigorous scientific and clinical validation can AM zirconia fulfill its potential to redefine the fabrication of dental restorations.

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