International Journal of Scientific Advances
ISSN: 2708-7972
Volume: 7 | Issue: 6 | Nov – Dec 2026 Available Online: www.ijscia.com
DOI: 10.51542/ijscia.v7i6.19

3D-Printed versus Conventionally Fabricated Scaffolds for Alveolar Bone Regeneration: A Narrative Review of Architecture, Mechanical Properties, and Regenerative Outcomes

Michael Josef Kridanto Kamadjaja[1*], Mefina Kuntjoro[1], Jesslyn Azalia[2]

[1]Department of Prosthodontic, Faculty of Dental Medicine,
Universitas Airlangga, Surabaya, Indonesia

[2]Undergraduate Student, Faculty of Dental Medicine,
Universitas Airlangga, Surabaya, Indonesia

E-mail: michael-j-k-k@fkg.unair.ac.id; mefina-k@fkg.unair.ac.id;
jesslyn.azalia-2023@fkg.unair.ac.id

*Corresponding author details: Michael Josef Kridanto Kamadjaja;
michael-j-k-k@fkg.unair.ac.id

ABSTRACT

Background: Alveolar ridge deficiency frequently requires bone augmentation before dental implant placement, and scaffold-based grafts are increasingly used as alternatives to autogenous bone. Conventional fabrication methods such as salt leaching, gas foaming, and freeze-drying yield porous scaffolds with limited control over internal architecture, whereas three-dimensional (3D) printing allows pore geometry and external shape to be predefined. ***Purpose:*** This narrative review compares 3D-printed and conventionally fabricated scaffolds for alveolar bone regeneration in terms of architecture, mechanical properties, and regenerative outcomes. ***Review:*** Nineteen articles published between 2016 and 2026 were reviewed, comprising six reviews, six *in vitro* or animal studies, and seven clinical reports. In same-material comparisons, printed scaffolds showed more regular and reproducible pore networks, higher compressive strength (4.0 vs 1.7 MPa), 21-fold higher alkaline phosphatase activity, and higher cell viability (about 85% vs 50%) than porogen-leached or freeze-dried scaffolds, although the strength difference coincided with lower porosity and one freeze-dried scaffold was stiffer than its printed counterparts. Printed polymer scaffolds also degraded more slowly *in vivo*. In randomized trials, 3D-printed ceramic grafts achieved new bone formation and implant stability comparable to non-printed grafts (33.2% vs 31.4% new bone), with a modest advantage in percent bone volume (20.0% vs 16.4%) for patient-customized blocks. The evidence remains limited by heterogeneous materials and designs, small samples, short follow-up, and few head-to-head trials. ***Conclusion:*** 3D printing offers superior architectural control, customization, and handling, but current clinical evidence indicates equivalence rather than clear superiority in bone regeneration over conventionally fabricated scaffolds.

Keywords: 3D printing; additive manufacturing; alveolar bone regeneration; bone tissue engineering; scaffold

INTRODUCTION

The success of dental implant therapy depends on an adequate volume of alveolar bone at the implant site. Ridge deficiencies resulting from tooth loss, periodontal disease, or trauma are frequently encountered and often necessitate horizontal or vertical bone augmentation to allow prosthetically driven implant positioning [1,2]. Bone remodeling after extraction commonly produces horizontal and vertical ridge defects that complicate implant placement [3]. Autogenous bone is still regarded as the gold standard because it is osteogenic, osteoinductive, and osteoconductive, yet its use is restricted by limited intraoral donor volume, donor-site morbidity, and the difficulty of adapting a harvested block to the recipient bed [4,5]. Conventional allografts, xenografts, and alloplasts avoid a second surgical site, but they are brittle, poorly processable into porous forms, and cannot be tailored to the needs of an individual defect [2].

Bone tissue engineering addresses these shortcomings through scaffolds that act as temporary three-dimensional templates for cell attachment, vascular ingrowth, and deposition of new bone matrix. An ideal scaffold should be biocompatible and resorbable at a rate matching bone repair, highly porous with an interconnected pore network, and mechanically comparable to the tissue it replaces [2,6]. Because these properties are largely determined by how the scaffold is made, the fabrication route is as important as the choice of biomaterial [7].

Conventional techniques, including salt leaching, gas foaming, phase separation, freeze-drying, and electrospinning, can generate porous structures but offer restricted control over pore architecture and limited ability to customize the scaffold for a particular defect [8]. Additive manufacturing, or three-dimensional (3D) printing, builds the scaffold layer by layer from a computer-aided design, so that the external shape and the internal pore network can be defined in advance and reproduced [8,9]. Combined with cone beam computed tomography, this enables patient-specific scaffolds for alveolar ridge augmentation [10].

Several reviews have summarized 3D-printed biomaterials, digital workflows, and resorbable scaffolds for alveolar ridge augmentation [1,2,10]. However, evidence that directly contrasts 3D-printed with conventionally fabricated scaffolds is scattered across *in vitro*, animal, and clinical studies that use different materials, designs, and outcome measures. This narrative review was therefore conducted to compare 3D-printed and conventionally fabricated scaffolds for alveolar bone regeneration with respect to architecture, mechanical properties, and regenerative outcomes. It draws on 19 articles published between 2016 and 2026, comprising six review articles, six *in vitro* or animal studies, and seven clinical reports, covering the topics of 3D printing or additive manufacturing, conventional scaffold fabrication, bone tissue engineering, and alveolar ridge augmentation or preservation.

DISCUSSION

Architectural requirements of scaffolds for alveolar bone

Native bone is a graded structure. Cortical bone is compact, with less than 10% porosity, whereas cancellous bone has a porosity of 50–90% and pore diameters of 100–1000 µm that provide channels for nutrient transport and metabolic activity [11]. Scaffolds are designed to reproduce the cancellous range: a total porosity of 30–90% is considered suitable for bone regeneration, and about 70% has been applied in preclinical and clinical alveolar studies, because excessive porosity reduces compressive strength [2]. Pore diameters of 150–500 µm facilitate vascularization and tissue penetration without compromising strength [2], and an interconnected porosity of 200–500 µm has been described as the ideal range for calcium phosphate blocks [4]. Pores smaller than 100 µm may lead to non-mineralized osteoid or fibrous tissue, so macropores of at least 400 µm with interconnections larger than 100 µm have been recommended [12].

Pore size, however, is only one level of the hierarchy. Macropores above 300 µm support cell migration and nutrient exchange, while micropores below 10 µm enhance osteoinductive capacity, and scaffolds combining both scales show better osseointegration than single-scale designs [9]. Surface roughness, hydrophilicity, and micro- to nanoscale topography further influence cell adhesion [2]. For dentoalveolar reconstruction, degradation within approximately 5–6 months is considered appropriate so that resorption keeps pace with remodeling [2]. These requirements form the benchmark against which the two fabrication routes can be compared.

Conventional fabrication techniques

Conventional methods create porosity indirectly. In salt leaching, a polymer solution is cast around a porogen that is later dissolved; porosity and pore size can be adjusted through the amount and size of the salt particles, but pore distribution and scaffold shape cannot be controlled, and residual organic solvent may impair cell viability [8]. Gas foaming avoids solvents, yet only 10–30% of the pores may be interconnected. Phase separation yields small pore diameters, freeze-drying is associated with long processing times and small irregular pores, and electrospinning produces pores of only a few tens of micrometres and is poorly suited to complex geometry [8]. Overall, these methods do not allow accurate regulation of the internal design, depend on operator skill for consistency, and tend to give poor mechanical properties [8]. Heterogeneity in pore size, porosity, and interconnectivity is unavoidable, and a custom-made scaffold that conforms precisely to a defect cannot readily be produced [2].

These techniques nevertheless retain real advantages. They are simple and can reach very high porosity, and they replicate micro- and nanoscale features more easily than current printers can [9]. In a direct comparison using the same poly(lactic-co-glycolic acid)/β-tricalcium phosphate (PLGA/β-TCP) composite, porogen-leached scaffolds reached 85% porosity against 39% for the printed design [13].

Additive manufacturing techniques and digital workflow

Additive manufacturing techniques are commonly grouped into material extrusion, vat photopolymerization, powder bed fusion, binder jetting, material jetting, sheet lamination, and directed energy deposition [9]. Those most often used for bone scaffolds are binder jetting, fused deposition modeling, selective laser sintering, and stereolithography [8]. The reviewed studies reflect this variety. Thermoplastic polymers and polymer–ceramic composites were printed by extrusion, such as fused filament fabrication of polycaprolactone /hydroxyapatite (PCL/HA) [7] and a medical-grade PCL ridge scaffold [14]. Calcium phosphate ceramics were produced by digital light processing [11], lithography-based ceramic manufacturing [5], dispense plotting of ceramic slurry [6], and powder-based printing [15], usually followed by sintering or another post-treatment.

For alveolar applications, the workflow begins with a cone beam computed tomography scan that is converted from DICOM to STL format, followed by virtual design of the scaffold, printing, and sterilization [10]. This allows a graft to be planned around the restoratively driven implant position [14]. 3D printing also has limitations. Extrusion printing is largely restricted to thermoplastic materials, and each technique differs in resolution [2,8].

Current printers struggle to reproduce features below about 100 µm, which can leave poorly defined pore boundaries [9]. Many processes require post-processing, and sintering shrinkage must be compensated in the design [8,11]. Ceramic loading also has an upper limit: in PCL/HA filaments the HA content was restricted to 20%, because higher loading reduced flowability and was expected to yield brittle, unprintable filaments [7].

Architecture: comparative evidence

Studies that processed one material by both routes give the clearest picture of how fabrication shapes architecture. Zamani et al. [13] compared PLGA/β-TCP scaffolds made by solvent casting/porogen leaching with printed scaffolds. The porous scaffolds had pores of 408 ± 90 µm and a highly rough, irregular surface (Ra 110 ± 15 µm), whereas the printed scaffolds had more uniform pores of 315 ± 17 µm, a smoother surface (Ra 22 ± 3 µm), and a lower water contact angle (76° vs 112°). The β-TCP particles were also distributed more homogeneously on the printed struts, while aggregates were seen on the porous scaffolds. Kwon et al. [16] reported that salt-leached poly(ε-caprolactone-ran-lactic acid) scaffolds had an irregular interconnected structure whose exact pore architecture could not be adjusted, whereas printed scaffolds had a layered, orthogonally oriented structure. Printing took about 3 minutes, compared with more than 3 days for salt removal and drying, and was judged superior in productivity and repeatability.

Pérez-Bohórquez et al. [3] fabricated HA/PLGA /chitosan scaffolds by freeze-drying and by 3D printing at 50%, 60%, and 70% infill. The freeze-dried scaffolds were heterogeneous and predominantly microporous, with irregular pores mostly below 3.5 µm and limited interconnectivity. The printed scaffolds showed hierarchical porosity that could be tuned through infill density, combining macropores of up to about 500 µm with micropores at 50% infill and becoming progressively more uniform at higher infill. Surface analysis also indicated greater calcium exposure on the printed scaffolds, although this was based on localized measurements.

3D printing further allows architectures that conventional methods cannot produce. Wang et al. [11] printed biphasic calcium phosphate scaffolds with a gyroid design in which peripheral pores of 400 µm surrounded central pores of 400–1000 µm at a constant porosity of 70%. Systermans et al. [12] used HA blocks with a fully interconnected gyroid porosity, noting that earlier synthetic HA blocks, with pores of 50–300 µm and interconnections of about 40 µm, often failed to osseointegrate because of inadequate porosity. In the clinical trial by Kim et al. [17], a conventional HA/β-TCP block had pores of 200–400 µm and about 80% porosity, whereas the printed patient-customized graft of similar composition had adjustable pores of 0.7–1.2 mm and 70–80% porosity.

Mechanical properties

Mechanical competence matters in ridge augmentation because the scaffold must maintain space under soft tissue pressure and tolerate fixation screws and implant drilling. In the PLGA/β-TCP comparison, the compressive strength of printed scaffolds was 4.0 ± 0.3 MPa, significantly higher than the 1.7 ± 0.2 MPa of porogen-leached scaffolds [13]. This difference should be interpreted with caution, since the printed scaffolds were far less porous (39% vs 85%) and compressive strength falls as porosity rises, as the authors themselves noted [13]. The finding therefore reflects freedom of design as much as an intrinsic advantage of the printed material.

Printing parameters and composition allow the mechanical response to be tuned. In printed PCL/HA scaffolds with 400 µm pores, Young's modulus under compression increased from 31.5 ± 2.0 MPa for pure PCL to 62.7 ± 10.6 MPa with 20% HA, within the 20–500 MPa range cited for cancellous bone [7]. Architecture can be changed while strength is preserved: at equal porosity, uniform 400 µm scaffolds and graded 400–600 and 400–800 µm scaffolds had similar compressive strengths (4.17, 3.99, and 3.71 MPa), whereas strength dropped to 2.46 MPa with 1000 µm central pores [11].

Printed scaffolds are not always stronger. Among the HA/PLGA/chitosan scaffolds, the highest apparent stiffness (40 MPa) belonged to a freeze-dried scaffold, while printed scaffolds ranged from 6 to 19 MPa, consistent with their more open architecture; finite element analysis under simulated implant loading indicated that further mechanical optimization was needed [3]. More broadly, measures that raise strength, such as high ceramic loading, dense design, and sintering, increase brittleness or reduce the surface available to cells, whereas very high porosity compromises load-bearing capacity [9]. In vivo, printed gyroid HA blocks could be fixed with screws and drilled with a standard implant protocol without fracture [12], and a printed PCL scaffold retained a particulate graft and maintained space during staged ridge augmentation [14].

Biological response and degradation

In vitro, pre-osteoblast seeding efficiency, proliferation, and collagen deposition were similar on printed and porogen-leached PLGA/β-TCP scaffolds, but alkaline phosphatase (ALP) activity was 21-fold higher on the printed scaffolds after 14 days. The authors attributed this to the moderate surface roughness, the homogeneous β-TCP distribution, and more effective cell–cell contact along the ordered struts [13]. With human dental pulp stem cells, printed HA/PLGA/chitosan scaffolds supported higher viability than freeze-dried scaffolds, about 85% at 50% infill compared with 50%, and the 70% infill design produced the strongest expression of RUNX2 and OSX [3]. Thus, the architecture that favours cell survival is not necessarily the one that favours osteogenic commitment. Adding HA to printed PCL also increased ALP activity, calcium deposition, and apatite formation in simulated body fluid [7].

The fabrication route also affects degradation and the host response. After subcutaneous implantation, fluorescence imaging showed that salt-leached scaffolds had degraded by 68% at 16 weeks, compared with 49% for printed scaffolds of the same polymer, a difference ascribed to rapid bulk erosion of the irregular porous structure [16]. The printed scaffolds, which persisted longer, showed slightly more macrophage infiltration and fewer fibrous capsules and blood vessels than the salt-leached scaffolds. Slower resorption is therefore not uniformly beneficial, and degradation time has been identified as a key but insufficiently studied determinant of clinical performance [1].

Pore design is the main lever for vascularization. Graded scaffolds with larger central pores increased permeability from 12.10 to 22.15 × 10^{-9} m^2, and the 400–800 µm design enhanced osteogenic differentiation *in vitro* and neovascularization and new bone formation after subcutaneous implantation in mice [11]. A systematic review of ten rodent calvarial studies found that pore sizes of at least 400 µm in printed composite scaffolds were associated with a higher bone volume fraction, and composite scaffolds with a larger new bone area than comparators, although these analyses were exploratory [18]. In minipigs, printed gyroid HA blocks used for lateral mandibular augmentation yielded 39.17% new bone and a mean bone gain of 4.73 mm, and implants placed through the blocks showed bone-to-implant contact similar to that in native bone (42.79% vs 34.17%) [12].

Clinical regenerative outcomes

Three clinical studies compared a 3D-printed graft with a non-printed counterpart (Table 1). In a randomized trial of alveolar ridge preservation in 30 patients, Kijartorn et al. [15] compared printed nano-porous HA granules with a commercial nano-crystalline bone graft. After four months, new bone formation (33.20% vs 31.35%), residual graft (27.04% vs 31.69%), dimensional bone changes, and primary implant stability did not differ significantly. Kim et al. [17] randomized 60 patients requiring guided bone regeneration to a printed patient-customized HA/β-TCP graft or a conventional block of similar composition. About five months later, percent bone volume was higher in the printed group (20.01% vs 16.35%; p = 0.049), but bone volume, bone surface, and bone mineral density were similar. The authors concluded that the customized graft simplified the procedure without demonstrating greater regenerative capability. In maxillary sinus augmentation, Giuliani et al. [6] compared rapid-prototyped HA/β-TCP blocks with granules of the same composition. Newly formed bone was similar at 5–6 months (26.3% vs 24.1%) and at 9 months (33.1% vs 30.1%), although the block specimens resembled native bone slightly more closely at 9 months.

Other clinical reports describe 3D-printed scaffolds without a conventionally fabricated comparator (Table 2). A patient-specific PCL scaffold filled with autogenous bone and bovine bone mineral produced a volumetric gain of 364.69 mm^3 and allowed implant placement with a primary stability of 35 Ncm [14]. Printed calcium phosphate blocks, manually shaped and covered with xenograft particles and a collagen membrane, showed ongoing bone regeneration and sprouting capillaries after six months in two patients [4]. Printed polymer–ceramic disks improved soft tissue closure of posterior extraction sockets compared with open healing [19]. A split-mouth randomized trial comparing a printed patient-specific β-TCP block with autogenous block grafts in the atrophic maxilla has been registered, but only its protocol is available [5]. A systematic review of resorbable scaffolds for ridge augmentation identified only seven eligible clinical studies, with 39 patients and 45 ridges, graft loss in five cases, and a mean primary implant stability of 35 Ncm [1]. Customized meshes and scaffolds have also been associated with improved surgical precision and shorter operative time in several studies [10].

TABLE 1: Studies Directly Comparing 3D-Printed and Conventionally Fabricated Scaffolds or Grafts.

Study	Material and model	3D-printed scaffold	Conventional comparator	Key comparative outcomes
Zamani et al. [13] (2021)	PLGA/β-TCP; *in vitro*, MC3T3-E1 pre-osteoblasts, 14 days	Extrusion-printed; pores 315 ± 17 µm; porosity 39%; Ra 22 µm	Solvent casting/porogen leaching; pores 408 ± 90 µm; porosity 85%; Ra 110 µm	Compressive strength 4.0 vs 1.7 MPa; ALP activity 21-fold higher on printed scaffolds; proliferation and collagen deposition similar
Kwon et al. [16] (2020)	Poly(ε-caprolactone-ran-lactic acid); subcutaneous implantation in rats and mice, 16 weeks	Fused deposition modeling; orthogonal layered strands; fabricated in about 3 min	Salt leaching (NaCl 200–250 µm); irregular interconnected pores; more than 3 days to complete	Degradation at 16 weeks 49% (printed) vs 68% (salt-leached); slightly more macrophages and fewer vessels around printed scaffolds
Pérez-Bohórquez et al. [3] (2026)	HA/PLGA/chitosan; *in vitro*, human dental pulp stem cells; finite element analysis	Extrusion-printed at 50%, 60%, and 70% infill; hierarchical pores up to about 500 µm	Freeze-drying; irregular micropores mostly below 3.5 µm; limited interconnectivity	Cell viability about 85% (50% infill) vs 50%; strongest RUNX2 and OSX expression at 70% infill; stiffness 6–19 MPa (printed) vs up to 40 MPa (freeze-dried)

Study	Material and model	3D-printed scaffold	Conventional comparator	Key comparative outcomes
Kijartorn et al. [15] (2022)	Hydroxyapatite; randomized trial of ridge preservation, 30 patients, 4 months	Printed nano-porous HA granules (1 mm)	Commercial nano-crystalline bone graft granules (0.6 mm)	New bone 33.20% vs 31.35%; residual graft 27.04% vs 31.69%; implant stability about 70 in both groups; no significant differences
Kim et al. [17] (2024)	HA/β-TCP 60:40; randomized trial of guided bone regeneration, 60 patients, about 5 months	Patient-customized printed graft; pores 0.7–1.2 mm; porosity 70–80%	Conventional block; pores 200–400 μm; porosity about 80%	BV/TV 20.01% vs 16.35% (p = 0.049); bone volume, bone surface, and bone mineral density not significantly different
Giuliani et al. [6] (2016)	HA/β-TCP 30/70; maxillary sinus augmentation, 24 sinuses, 5–6 and 9 months	Blocks made by dispense plotting (direct rapid prototyping)	Granules of the same composition	New bone 26.3% vs 24.1% at 5–6 months and 33.1% vs 30.1% at 9 months (not significant); blocks slightly closer to native bone at 9 months

ALP, alkaline phosphatase; BV/TV, percent bone volume; HA, hydroxyapatite; PLGA, poly(lactic-co-glycolic acid); Ra, surface roughness; β-TCP, β-tricalcium phosphate. Values are given as 3D-printed vs conventional.

TABLE 2: Studies of 3D-Printed Scaffolds for Alveolar Bone Augmentation without a Conventionally Fabricated Scaffold Comparator.

Study	Scaffold and printing method	Model and application	Key reported outcomes
Systermans et al. [12] (2026)	Pure HA blocks with interconnected gyroid porosity	Four minipigs; bilateral lateral mandibular augmentation; implant placed after 8 or 12 weeks	39.17% new bone; mean bone gain 4.73 mm; bone-to-implant contact 42.79% in the block vs 34.17% in native bone; no control group
Ivanovski et al. [14] (2024)	Patient-specific medical-grade PCL scaffold printed by extrusion; filled with autogenous bone and bovine bone mineral	Case report; staged horizontal augmentation at a maxillary central incisor site; 6 months	Volumetric bone gain 364.69 mm^3; implant primary stability 35 Ncm; new bone in all defect zones
Pérez et al. [4] (2023)	Printed calcium phosphate blocks, manually shaped; covered with xenograft particles and collagen membrane	Case series of two patients; staged horizontal ridge augmentation; 6 months	Implants placed; histology showed ongoing bone regeneration and sprouting capillaries
De Angelis et al. [19] (2024)	Printed disks of poly-D-lactic acid with 10% HA or PCL with 20% β-TCP	Pilot randomized study; 39 patients; socket sealing vs open healing; 6 weeks	Posterior sockets: exposed area reduced to 0.6 mm^2 in both test groups vs 1.2 mm^2 in controls ($p < 0.05$); soft tissue outcome only
de Almeida Malzoni et al. [5] (2022)	Patient-specific β-TCP block made by lithography-based ceramic manufacturing	Split-mouth randomized trial protocol; atrophic anterior maxilla; autogenous block as control; 8 months	Protocol only; volumetric and histological outcomes not yet reported

HA, hydroxyapatite; PCL, polycaprolactone; β-TCP, β-tricalcium phosphate.

Synthesis of the comparison

Taken together, the reviewed studies point to a consistent pattern. The clearest advantage of 3D printing lies in architecture: pore size, strut orientation, gradients, and external shape can be specified and reproduced, whereas conventionally fabricated scaffolds are irregular and vary between batches [3,13,16]. Mechanical properties follow from this design freedom and are not inherently superior, since they depend on the porosity and composition chosen [3,11,13]. Biologically, ordered architecture favoured early osteogenic markers *in vitro* [3,13], but this has not yet translated into clearly greater bone formation in patients, where printed and non-printed ceramic grafts performed similarly [6,15,17]. The practical benefits reported clinically relate mainly to fit, handling, and procedural simplicity [10,17].

Limitations and challenges

Several limitations restrict the conclusions that can be drawn. First, the comparisons are confounded. Fabrication route was rarely the only variable, because porosity, pore size, surface roughness, and even the formulation differed between the printed and conventional groups [3,13], and in the clinical comparisons, granule size or graft form differed as well [6,15]. Second, much of the comparative evidence comes from short-term *in vitro* work or ectopic subcutaneous models rather than alveolar

defects [3,11,13,16]. Third, the clinical evidence is sparse: it consists largely of case reports, pilot studies, and a protocol [4,5,14,19]; follow-up rarely exceeded nine months, and the only minipig study had a small sample, no control group, and an extraoral approach that avoids the wound dehiscence seen intraorally [12]. Heterogeneity in materials, surgical protocols, and outcome reporting has so far precluded meta-analysis [1,18].

Technical and translational barriers also remain. Printing resolution limits the reproduction of microscale features [9], shrinkage and post-processing affect dimensional accuracy [8,11], and the slower degradation of dense printed struts may prolong the foreign body response [16]. Regulatory pathways for 3D-printed scaffolds are complex, and the customization and low scalability of printed constructs increase cost [9]. Variability in digital protocols and sterilization methods further complicates standardization [10].

Clinical implications and future directions

From a clinical perspective, 3D-printed scaffolds are most attractive where conventional grafts are weakest: large or complex defects in which a block must fit the recipient bed closely and maintain volume [5,12]. A patient-specific scaffold can be planned around the intended implant position, may shorten surgery, and avoids a donor site [10,14]. For contained defects such as extraction sockets, however, conventional particulate grafts achieve comparable bone formation [15], and the additional planning and cost of printing may not be justified. Printed scaffolds should therefore be regarded as a complementary option selected according to defect morphology, not as a replacement for established materials.

Conventionally fabricated scaffolds are also being developed further, beyond the 19 articles reviewed here. At the authors' institution, lyophilized chitosan–hydroxyapatite scaffolds have been studied as biologically active options for bone regeneration: a scaffold seeded with hypoxic human amnion mesenchymal stem cells increased angiogenesis, osteogenic marker expression, and trabecular bone area in rat calvarial defects compared with untreated controls [20], and a freeze-dried scaffold seeded with cryopreserved adipose-derived stem cells was associated with significantly higher bone sialoprotein secretion on day 14 [21]. A systematic review of 15 studies likewise concluded that three-dimensional chitosan–hydroxyapatite scaffolds provide room for new bone formation [22]. The scaffolds in the two experimental studies were lyophilized rather than printed, so they are best regarded as a conventional benchmark against which printed scaffolds should be compared in future studies.

The two approaches are also increasingly combined. Hybrid strategies that use printing for the macroscale framework and freeze-drying, gas foaming, or electrospinning for micro- and nanoscale features aim to reproduce the multiscale hierarchy of bone more faithfully than either route alone [9]. Future studies should compare printed and conventionally fabricated scaffolds of identical composition and matched porosity, test them in alveolar defect models and adequately powered randomized trials, and report standardized outcomes, including volumetric bone gain, histomorphometry, complication categories, implant stability and survival, and scaffold degradation over longer follow-up [1,18].

CONCLUSIONS

3D printing provides a level of control over scaffold architecture that conventional fabrication techniques cannot match, producing regular, reproducible, and patient-specific pore networks, including graded and gyroid designs. This control allows mechanical properties to be tuned and has been associated with stronger early osteogenic responses *in vitro* and slower degradation *in vivo* than porogen-leached or freeze-dried scaffolds of the same material. Conventional methods remain simple, reach high porosity, and reproduce microscale features that current printers cannot. In clinical trials, 3D-printed ceramic grafts have achieved bone formation and implant stability comparable to non-printed grafts, with advantages mainly in fit and surgical convenience. Given the confounded comparisons, small samples, and short follow-up of the available studies, 3D-printed scaffolds should currently be considered a promising, defect-specific alternative whose regenerative superiority is not yet established. Well-controlled head-to-head trials with standardized outcomes are needed.

ACKNOWLEDGMENT

The authors would like to thank the Faculty of Dental Medicine, Universitas Airlangga, for the support provided during the preparation of this article.

REFERENCES

[1] Seiler, M., Cucchi, A., Tröltzsch, M., Tröltzsch, M., Kämmerer, P.W., Al-Nawas, B. and Hartmann, A. (2026). Resorbable scaffolds for alveolar ridge augmentation procedures: A systematic review. International Journal of Implant Dentistry, 12, 27. https://doi.org/10.1186/s40729-026-00679-1

[2] Asa'ad, F., Pagni, G., Pilipchuk, S.P., Giannì, A.B., Giannobile, W.V. and Rasperini, G. (2016). 3D-printed scaffolds and biomaterials: Review of alveolar bone augmentation and periodontal regeneration applications. International Journal of Dentistry, 2016, 1239842. https://doi.org/10.1155/2016/1239842

[3] Pérez-Bohórquez, J.M., Acero-Garzón, M.C., Gutiérrez-Prieto, S.J., Méndez-Pinzón, H.A., Perdomo-Lara, S.J., Rodríguez-Hernández, H., Solís-Valencia, M.B. and Sequeda-Castañeda, L.G. (2026). HAp/PLGA/chitosan scaffolds fabricated by freeze-drying and 3D printing for bone regeneration: *In vitro* evaluation and finite element analysis. Biomimetics, 11(8), 537. https://doi.org/10.3390/biomimetics11080537

[4] Perez, A., Lazzarotto, B., Marger, L. and Durual, S. (2023). Alveolar ridge augmentation with 3D-printed synthetic bone blocks: A clinical case series. Clinical Case Reports, 11, e7171. https://doi.org/10.1002/ccr3.7171

[5] de Almeida Malzoni, C.M., Gonçalves, V., Possari, J. and Marcantonio Junior, E. (2022). The use of 3D ceramic block graft compared with autogenous block graft for rehabilitation of the atrophic maxilla: a randomized controlled clinical trial. Trials, 23, 903. https://doi.org/10.1186/s13063-022-06843-3

[6] Giuliani, A., Manescu, A., Mohammadi, S., Mazzoni, S., Piattelli, A., Mangano, F., Iezzi, G. and Mangano, C. (2016). Quantitative kinetics evaluation of blocks versus granules of biphasic calcium phosphate scaffolds (HA/β-TCP 30/70) by synchrotron radiation X-ray microtomography: A human study. Implant Dentistry, 25(1), 6–15. https://doi.org/10.1097/ID.0000000000000363

[7] Rezania, N., Asadi-Eydivand, M., Abolfathi, N., Bonakdar, S., Mehrjoo, M. and Solati-Hashjin, M. (2022). Three-dimensional printing of polycaprolactone/hydroxyapatite bone tissue engineering scaffolds mechanical properties and biological behavior. Journal of Materials Science: Materials in Medicine, 33, 31. https://doi.org/10.1007/s10856-022-06653-8

[8] Mohammed, A., Jiménez, A., Bidare, P., Elshaer, A., Memic, A., Hassanin, H. and Essa, K. (2024). Review on engineering of bone scaffolds using conventional and additive manufacturing technologies. 3D Printing and Additive Manufacturing, 11(4), 1418–1440. https://doi.org/10.1089/3dp.2022.0360

[9] Zhu, Y., Gao, H., Qiao, Q., Yuan, Y., Fang, D., Bai, Y. and Jiang, Q. (2025). Integrating additive and traditional manufacturing for multiscale bone tissue engineering scaffolds. Journal of Functional Biomaterials, 16(9), 349. https://doi.org/10.3390/jfb16090349

[10] Elrefaei, S.A., Parma-Benfenati, L., Dabaja, R., Nava, P., Wang, H.-L. and Saleh, M.H.A. (2025). Customized 3D-printed scaffolds for alveolar ridge augmentation: A scoping review of workflows, technology, and materials. Medicina, 61(7), 1269. https://doi.org/10.3390/medicina61071269

[11] Wang, Y., Liu, Y., Chen, S., Siu, M.-F.F., Liu, C., Bai, J. and Wang, M. (2025). Enhancing bone regeneration through 3D printed biphasic calcium phosphate scaffolds featuring graded pore sizes. Bioactive Materials, 46, 21–36. https://doi.org/10.1016/j.bioactmat.2024.11.024

[12] Systermans, S., Ronsmans, C., Nolens, G., Meyns, J., Gilon, Y. and Cobraiville, E. (2026). Dental implant osseointegration following mandibular augmentation using 3D printed hydroxyapatite blocks. An experimental *in vivo* study. Clinical Oral Implants Research, 37, 101–113. https://doi.org/10.1111/clr.70056

[13] Zamani, Y., Amoabediny, G., Mohammadi, J., Zandieh-Doulabi, B., Klein-Nulend, J. and Helder, M.N. (2021). Increased osteogenic potential of pre-osteoblasts on three-dimensional printed scaffolds compared to porous scaffolds for bone regeneration. Iranian Biomedical Journal, 25(2), 78–87. https://doi.org/10.29252/ibj.25.2.78

[14] Ivanovski, S., Staples, R., Arora, H., Vaquette, C. and Alayan, J. (2024). Alveolar bone regeneration using a 3D-printed patient-specific resorbable scaffold for dental implant placement: A case report. Clinical Oral Implants Research, 35, 1655–1668. https://doi.org/10.1111/clr.14340

[15] Kijartorn, P., Wongpairojpanich, J., Thammarakcharoen, F., Suwanprateeb, J. and Buranawat, B. (2022). Clinical evaluation of 3D printed nano-porous hydroxyapatite bone graft for alveolar ridge preservation: A randomized controlled trial. Journal of Dental Sciences, 17(1), 194–203. https://doi.org/10.1016/j.jds.2021.05.003

[16] Kwon, D.Y., Park, J.Y., Lee, B.Y. and Kim, M.S. (2020). Comparison of scaffolds fabricated via 3D printing and salt leaching: *In vivo* imaging, biodegradation, and inflammation. Polymers, 12(10), 2210. https://doi.org/10.3390/polym12102210

[17] Kim, N.-H., Yang, B.-E., On, S.-W., Kwon, I.-J., Ahn, K.-M., Lee, J.-H. and Byun, S.-H. (2024). Customized three-dimensional printed ceramic bone grafts for osseous defects: a prospective randomized study. Scientific Reports, 14, 3397. https://doi.org/10.1038/s41598-024-53686-w

[18] Vitosyte, M., Tesing, M., Galinauskaite, S., Rutkunas, V. and Gendviliene, I. (2026). Vascularization and bone regeneration with 3D-printed composite scaffolds in rodent critical-size calvarial defects: Systematic review. Journal of Functional Biomaterials, 17(3), 115. https://doi.org/10.3390/jfb17030115

[19] De Angelis, N., Pesce, P., Poedjiastoeti, W., Suwandi, T., Tjandrawinata, R., Bagnasco, F. and Menini, M. (2024). 3D-printable biopolymers for socket preservation technique: Soft tissues response: A pilot randomised clinical study. Dentistry Journal, 12(10), 321. https://doi.org/10.3390/dj12100321

[20] Kamadjaja, M.J.K. (2021). Bone remodeling using a three-dimensional chitosan–hydroxyapatite scaffold seeded with hypoxic conditioned human amnion mesenchymal stem cells. Dental Journal (Majalah Kedokteran Gigi), 54(2), 68–73. https://doi.org/10.20473/j.djmkg.v54.i2.p68-73

[21] Kamadjaja, M.J.K., Kuntjoro, M., Pratiwi, C.B. and Hartanli, A.L. (2025). Effect of chitosan–hydroxyapatite scaffolds seeded with cryopreserved hADMSCs on bone sialoprotein secretion. Majalah Kedokteran Gigi Indonesia, 11(3). https://doi.org/10.22146/majkedgiind.110209

[22] Kamadjaja, M.J.K., Laksono, H., Masirri, V.J. and Safitri, E.D. (2022). *In vitro* and *in vivo* examination of the osteogenic ability of 3D scaffold chitosan-hydroxyapatite: a systematic review. Indonesian Journal of Prosthodontics, 3(1), 6–12.