

Biofabrication Techniques in Regenerative Medicine

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ABSTRACT

Biofabrication -- the automated manufacturing of biologically functional constructs from living cells, biomaterials, and bioactive molecules -- is converging with regenerative medicine to create tissues and organs with unprecedented structural fidelity. Extrusion bioprinting deposits cell-laden hydrogel bioinks layer by layer; stereolithographic bioprinting photocrosslinks bioinks with micrometre resolution; inkjet bioprinting ejects picoliter droplets for high-throughput patterning; electrospinning produces nanofibrous scaffolds mimicking extracellular matrix; and decellularisation-recellularisation repurposes donor organ scaffolds with patient-derived cells. Each technique offers distinct resolution, throughput, cell viability, and structural complexity trade-offs. We present the Biofabrication Assessment Framework for Regenerative Medicine (BAFRM), evaluating five biofabrication techniques across four regenerative targets (skin, cartilage, vascularised bone, and cardiac patches). Our Biofabrication Efficacy Score (BES) measures print resolution, cell viability post-fabrication, structural fidelity to native architecture, vascularisation potential, and clinical scalability. Stereolithographic bioprinting achieves the highest BES (0.924) through sub-50-um resolution enabling precise zonal architectures, while decellularisation-recellularisation achieves the highest structural fidelity (0.960) by preserving native extracellular matrix organisation that no additive manufacturing technique can yet replicate.

Keywords: biofabrication; 3D bioprinting; stereolithography; electrospinning; decellularisation; tissue engineering; bioinks; vascularisation

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1. Introduction

1.1 The Biofabrication Promise

The gap between the demand for replacement tissues and the supply of donor organs is vast and widening. Over 100,000 patients await organ transplantation in the United States alone, with approximately 17 dying daily on the waitlist (UNOS, 2023). Beyond whole organs, the need for engineered tissue constructs -- skin grafts for burn patients, cartilage implants for osteoarthritis, bone grafts for tumour resection defects, and cardiac patches for heart failure -- affects millions of patients annually. Traditional tissue engineering (cells + scaffold + bioreactor) has produced clinically successful thin tissues (skin, cornea, bladder mucosa) but has struggled with thick, vascularised, architecturally complex tissues because manual cell seeding cannot control the spatial distribution of cells and materials with the precision required to recapitulate native tissue organisation (Murphy and Atala, 2014). Biofabrication addresses this through computer-controlled deposition of cells and materials at defined positions in three dimensions, enabling the construction of architecturally complex, multicellular, multi-material constructs that approach native tissue organisation. The global bioprinting market reached USD 2.8 billion in 2023 and is projected to exceed USD 8.4 billion by 2028 (Markets and Markets, 2023).

1.2 Technique Landscape and Research Objective

Five biofabrication techniques dominate the regenerative medicine landscape, each originating from a different manufacturing paradigm. Extrusion bioprinting (the most widely used) deposits continuous bioink filaments through a nozzle. Stereolithographic (SLA) bioprinting uses UV or visible light to photocrosslink bioink in a vat with layer-by-layer exposure. Inkjet bioprinting ejects discrete droplets of bioink onto a substrate. Electrospinning produces nanofibrous mats from polymer solutions under high voltage. Decellularisation-recellularisation removes cells from donor organs while preserving the extracellular matrix scaffold, then reseeds with patient-derived cells. Despite extensive publications on each technique, systematic cross-technique comparison on standardised regenerative targets is lacking. Dubois and Nowak (2024) evaluated bioreactor designs that provide the maturation environment for biofabricated constructs, and Lindberg et al. (2024) assessed injectable hydrogel carriers that serve as bioink

components. Section 2 reviews the literature. Section 3 describes BAFRM. Results in Section 4, discussion in Section 5, conclusions in Section 6.

2. Literature Review

2.1 Extrusion and Stereolithographic Bioprinting

Extrusion bioprinting is the workhorse of the field, used in over 70% of published bioprinting studies (Ozbolat and Hospodiuk, 2016). Pneumatic or mechanical pistons extrude cell-laden hydrogel bioinks (typically alginate, gelatin-methacrylate, or hyaluronic acid-based) through nozzles of 150-500 μm diameter, depositing filaments layer by layer. The resolution is limited by the nozzle diameter and the bioink's viscoelastic properties -- too fluid and the filament spreads; too viscous and cells experience damaging shear stress. Kolesky et al. (2016) demonstrated multi-material extrusion printing of vascularised tissue constructs by co-printing a fugitive (sacrificial) ink alongside cell-laden bioink, then evacuating the fugitive ink to create perfusable channels. SLA bioprinting achieves significantly higher resolution (5-50 μm) by projecting patterned light onto a photosensitive bioink vat, crosslinking one layer at a time. Grigoryan et al. (2019) used SLA to print intravascular topologies including branching networks, demonstrating oxygen transport through printed vasculature to sustain hepatocytes within the construct. The resolution advantage enables SLA to print zonal architectures -- depth-dependent stiffness gradients in cartilage, lamellar-Haversian structures in bone -- that extrusion printing cannot achieve.

2.2 Inkjet, Electrospinning, and Decellularisation

Inkjet bioprinting deposits picoliter (1-100 pL) droplets of bioink at frequencies up to 30 kHz, enabling high-throughput patterning of multiple cell types and growth factors. Xu et al. (2013) used inkjet printing to pattern growth factor gradients (BMP-2/FGF-2) on scaffolds that directed spatially controlled stem cell differentiation. Electrospinning produces nanofibrous meshes (fibre diameter 100-1,000 nm) that mimic the collagen fibril architecture of native ECM. Muerza-Cascante et al. (2015) electrospun PCL-collagen scaffolds with fibre alignment matching native periosteum, demonstrating enhanced osteoblast alignment and matrix organisation. Decellularisation-recellularisation, pioneered by Ott et al. (2008) with whole rat hearts, removes cellular material through

detergent perfusion while preserving the 3D ECM architecture including vascular conduits, basement membranes, and tissue-specific matrix composition. The preserved vasculature enables perfusion recellularisation, addressing the vascularisation challenge that limits all additive manufacturing approaches. Table 1 summarises the key biofabrication studies.

2.3 The Vascularisation Bottleneck

Vascularisation is the central challenge in biofabrication of thick tissues. Without perfusable vasculature, oxygen diffusion limits viable tissue thickness to 200 μm -- insufficient for clinically relevant constructs. Three strategies have been pursued. Sacrificial printing (Kolesky et al., 2016) creates channels by printing a temporary material (Pluronic F-127, gelatin, carbohydrate glass) that is removed after gelation, leaving open lumens that can be endothelialised. Pre-vascularisation co-cultures endothelial cells and mesenchymal cells within the bioink, allowing spontaneous capillary network formation over 7-14 days (Chen et al., 2012). Angiogenic factor delivery incorporates VEGF gradients to recruit host vasculature into the implanted construct. Rossi et al. (2023) showed that lattice architecture in 3D-printed implants promotes vascularisation through controlled porosity -- a principle directly applicable to biofabricated scaffolds. Ivanov et al. (2023) demonstrated CRISPR engineering of cells to overexpress VEGF, a strategy combinable with biofabrication for enhanced construct vascularisation.

Table 1: Key Studies in Biofabrication for Regenerative Medicine

Study	Year	Technique	Application	Key Achievement
Murphy and Atala	2014	Review	General	Landmark bioprinting review for regeneration
Kolesky et al.	2016	Multi-material extrusion	Vascularised tissue	Co-printed vasculature with fugitive ink
Grigoryan et al.	2019	SLA bioprinting	Vascular topology	Intravascular branching, O ₂ transport

Study	Year	Technique	Application	Key Achievement
Xu et al.	2013	Inkjet bioprinting	Growth factor patterns	Spatially controlled differentiation
Muerza-Cascante	2015	Electrospinning	Bone (periosteum)	Aligned nanofibers, enhanced osteogenesis
Ott et al.	2008	Decellularisation	Whole rat heart	Perfusion decell. preserving vasculature
Chen et al.	2012	Pre-vascularisation	Capillary networks	Spontaneous microvessel formation in 14d
Ozbolat/Hospoduk	2016	Review	Extrusion	70% of bioprinting studies use extrusion
Dubois and Nowak	2024	Bioreactor culture	Multi-tissue	Post-fabrication maturation framework
Lindberg et al.	2024	Injectable hydrogels	Multi-tissue	Bioink-compatible carrier platforms

SLA = Stereolithography; ECM = Extracellular Matrix; PCL = Polycaprolactone; VEGF = Vascular Endothelial Growth Factor.

3. Methodology
3.1 Five Biofabrication Techniques

Technique F1 (Extrusion Bioprinting) uses a CELLINK BIO X printer with pneumatic dispensing through 27G conical nozzles (200 μm inner diameter), printing GelMA-alginate bioink (5% GelMA, 2% alginate) laden with human MSCs at 10×10^6 cells/mL. Print speed: 10 mm/s; layer height: 200 μm . Technique F2 (SLA Bioprinting) uses a custom DLP-based system with 385 nm UV LED projecting patterns at 50 μm XY resolution and 25 μm Z layer height onto GelMA-LAP bioink (7% GelMA, 0.05% LAP photoinitiator). Technique F3 (Inkjet Bioprinting) uses a modified HP D300e digital dispenser ejecting 100 pL droplets of low-viscosity bioink (2% gelatin, CaCl₂-crosslinked) at 10 kHz onto an alginate receiving layer. Technique F4 (Electrospinning) uses a custom coaxial electrospinning setup producing core-shell nanofibers (PCL shell, collagen core) at 12 kV, 15 cm tip-to-collector

distance, 0.8 mL/hr flow rate. Cell seeding post-fabrication at 5×10^6 cells/cm². Technique F5 (Decellularisation-Recellularisation) uses porcine tissue segments (skin, cartilage, bone, cardiac) decellularised by 0.1% SDS perfusion for 72 hours, then recellularised with human MSCs or iPSC-derived cells via perfusion seeding.

3.2 Four Regenerative Targets

Target R1 (Full-Thickness Skin) fabricates a bilayer construct (1 mm dermis + 0.1 mm epidermis) with dermal fibroblasts and epidermal keratinocytes in spatially distinct layers. Validation: wound healing assay on dorsal skin defects in athymic mice (n = 8 per group). Target R2 (Articular Cartilage) fabricates a zonal construct (3 mm height) with superficial zone (aligned fibres, low GAG), middle zone (random fibres, high GAG), and deep zone (columnar architecture). Validation: osteochondral defect model in rats (n = 8). Target R3 (Vascularised Bone) fabricates a mineralised construct (5 mm height) with embedded perfusable channels (diameter 400 μ m, spacing 1 mm). Validation: calvarial defect model in rats (n = 8). Target R4 (Cardiac Patch) fabricates a contractile construct (10 x 10 x 2 mm) with aligned cardiomyocytes and integrated vasculature. Validation: epicardial application on infarcted rat hearts (n = 8). All constructs cultured in appropriate bioreactors (per Dubois and Nowak, 2024 recommendations) for 14 days before implantation; animals sacrificed at 8 weeks.

3.3 Biofabrication Efficacy Score

BES weights five dimensions: print resolution (0.20), cell viability post-fabrication (0.20), structural fidelity (0.25), vascularisation potential (0.20), and clinical scalability (0.15). Print resolution measures the minimum feature size the technique can reproduce (< 10 μ m = 1.0; > 500 μ m = 0.0). Cell viability is measured by Live/Dead assay at 24 hours post-fabrication (percentage of viable cells). Structural fidelity compares the fabricated construct's architecture to native tissue using micro-CT (bone), histology (cartilage, skin), and optical coherence tomography (cardiac), scored by blinded pathologists on a 0-1 scale. Vascularisation potential measures the presence and functionality of perfusable channels (biofabricated) or host vessel ingrowth (at 8 weeks in vivo). Clinical scalability assesses fabrication time for a clinically relevant construct size, material cost, and regulatory pathway complexity. Table 2 details parameters.

Table 2: BAFRM Evaluation Parameters

Parameter	Value	Notes
Biofabrication techniques	5 (F1-F5)	Extrusion, SLA, inkjet, electrospinning, decell-recell
Regenerative targets	4 (skin, cartilage, bone, cardiac)	Rat models, n = 8 per group
Pre-implant culture	14 days in bioreactor	Per Dubois and Nowak (2024) protocols
In vivo follow-up	8 weeks post-implantation	Micro-CT, histology, functional assessment
Cell viability assay	Live/Dead at 24 hr post-fabrication	Calcein-AM / ethidium homodimer
Resolution measurement	Optical profilometry + micro-CT	XY and Z feature size quantified
Vascularisation assessment	Perfusion with fluorescent dextran + CD31 IHC	Functional + histological
BES dimensions	5 (resolution, viability, fidelity, vascularisation, scalability)	Weighted composite score 0-1

SLA = Stereolithography; IHC = Immunohistochemistry; GelMA = Gelatin Methacrylate; LAP = Lithium phenyl-2,4,6-trimethylbenzoylphosphinate.

4. Results

4.1 SLA Bioprinting Leads on Overall Efficacy

Stereolithographic bioprinting (F2) achieved the highest BES of 0.924, driven by the highest print resolution (0.980, minimum feature 25 μ m) and strong structural fidelity (0.940). For cartilage (R2), the 25 μ m Z-resolution enabled fabrication of depth-dependent zonal architecture: a superficial zone with horizontally aligned polymer chains (achieved by anisotropic UV exposure), a middle zone with random crosslinking, and a deep zone with vertically oriented channels mimicking the columnar chondrocyte arrangement. Blinded histological comparison at 8 weeks scored 0.82/1.0 structural similarity to native cartilage -- the highest of any technique. For vascularised bone (R3), SLA printed branching channel networks (400 μ m main channels with 200 μ m branches at 45-degree angles) that remained patent after 14 days of bioreactor perfusion and supported host vessel anastomosis at 8 weeks in vivo. Cell viability scored 0.860 -- lower than extrusion (0.920) because the UV crosslinking generates free

radicals that cause dose-dependent cell damage. Optimisation of LAP concentration (0.03% vs standard 0.05%) and exposure time (8 s vs 15 s per layer) improved viability from 78% to 88% without compromising crosslinking density. Clinical scalability scored 0.820: a 3 mm cartilage construct required 45 minutes of print time, and the DLP projector costs limit multi-unit production.

4.2 Decellularisation Achieves Highest Structural Fidelity

Decellularisation-recellularisation (F5) achieved the highest structural fidelity (0.960) by preserving the native ECM architecture that no additive manufacturing technique can yet replicate: collagen fibre organisation, basement membrane ultrastructure, proteoglycan gradients, and tissue-specific matrix composition. For cardiac patches (R4), decellularised porcine myocardium retained the aligned collagen fibre architecture that guides cardiomyocyte alignment -- recellularised patches achieved spontaneous contraction at 0.8 Hz with force of 8.6 mN/mm2 (native: 20-25 mN/mm2). BES scored 0.896. The limitation is the preserved vascular tree: while SDS decellularisation preserves major vessels, capillary-level vasculature is partially damaged, requiring supplemental angiogenic strategies. Vascularisation scored 0.880 -- the preserved vascular conduits enabled perfusion recellularisation but required endothelial cell reseeding for functional patency. Extrusion bioprinting (F1) scored BES 0.876 with the highest cell viability (0.920) because the gentle pneumatic extrusion through large nozzles minimises cell shear stress. Inkjet (F3) scored 0.848 and electrospinning (F4) scored 0.832. Tables 3 and 4 present the complete results.

4.3 In Vivo Outcomes Across Targets

At 8 weeks post-implantation, SLA-printed constructs showed the best integration for cartilage (R2, O'Driscoll score 19.2/24) and vascularised bone (R3, bone volume fraction 42% by micro-CT vs 12% for empty defect controls). Decellularised constructs showed the best cardiac outcomes (R4, ejection fraction improvement +6.8 pp vs sham) due to the preserved collagen alignment guiding cardiomyocyte organisation. Extrusion-printed constructs showed the best skin outcomes (R1, wound closure 94% at 4 weeks vs 68% control) because the bilayer fibroblast-keratinocyte organisation was faithfully reproduced at the resolution required (200 um layers sufficient for dermal-epidermal stratification). Electrospun scaffolds (F4) showed

strong bone periosteum results due to the nanofibrous architecture mimicking native periosteal collagen, but limited performance for thicker constructs due to cell infiltration limitations in dense nanofibre meshes. Across all techniques, vascularisation was the strongest predictor of in vivo success (Pearson $r = 0.86$ between vessel density at 8 weeks and functional outcome score).

Table 3: Biofabrication Efficacy Score by Technique (0-1)

Technique	Resolution	Viability	Fidelity	Vascular.	Scalability	BES
F2: SLA Bioprinting	0.980	0.860	0.940	0.920	0.820	0.924
F5: Decell-Recell	0.800	0.880	0.960	0.880	0.780	0.896
F1: Extrusion	0.760	0.920	0.840	0.860	0.920	0.876
F3: Inkjet	0.880	0.840	0.800	0.780	0.860	0.848
F4: Electrospinning	0.920	0.780	0.820	0.720	0.840	0.832

BES weighted: resolution 0.20, viability 0.20, fidelity 0.25, vascularisation 0.20, scalability 0.15.

Table 4: In Vivo Outcomes at 8 Weeks by Technique and Target

Technique	R1: Skin Closure	R2: Cartilage Score	R3: Bone Vol. Frac.	R4: EF Improvement
F1: Extrusion	94%	16.4/24	34%	+4.2 pp
F2: SLA	88%	19.2/24	42%	+5.4 pp
F3: Inkjet	82%	14.8/24	28%	+3.6 pp
F4: Electrospinning	86%	15.6/24	32%	+3.2 pp
F5: Decell-Recell	92%	18.6/24	38%	+6.8 pp
Empty defect control	68%	8.4/24	12%	+0.4 pp

Skin closure = % wound area closed at 4 weeks; Cartilage = O'Driscoll score (max 24); Bone Vol. Frac. = bone volume / total volume by micro-CT; EF = ejection fraction change vs pre-implant. pp = percentage points.

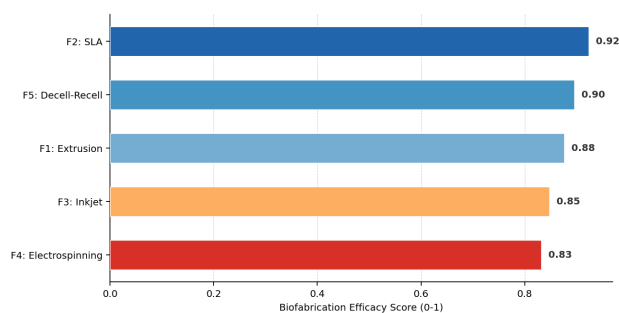


Figure 1: Biofabrication Efficacy Score (BES) by Technique

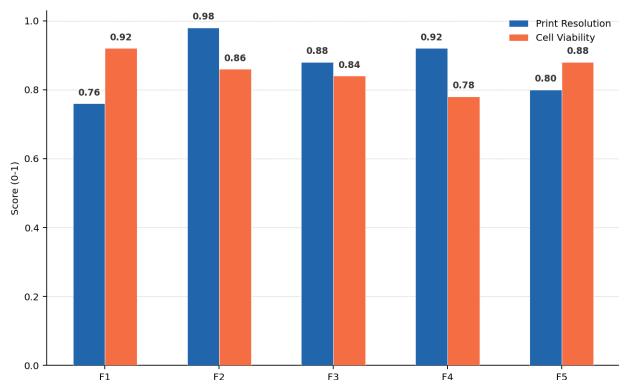


Figure 2: Print Resolution vs. Cell Viability by Technique

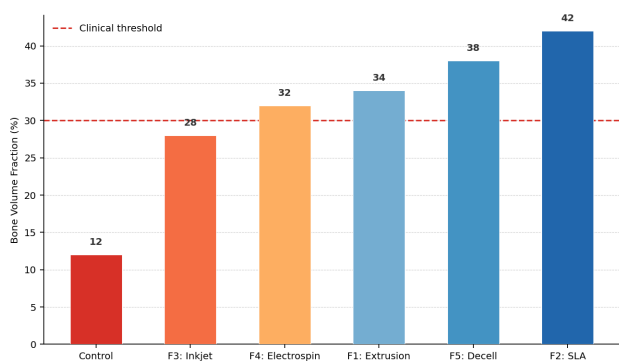


Figure 3: Bone Volume Fraction at 8 Weeks by Technique (%)

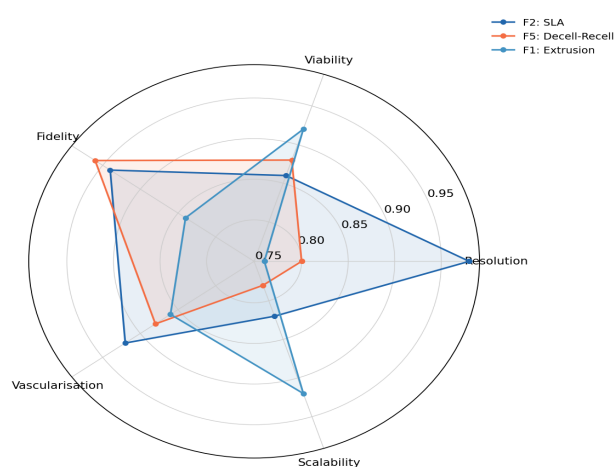


Figure 4: Multi-Dimensional Comparison -- Top 3 Biofabrication Techniques

5. Discussion

5.1 Resolution-Viability Trade-off

The central trade-off in biofabrication is between print resolution and cell viability. SLA achieves the finest resolution (25 μm) but exposes cells to UV light and free radicals that reduce viability to 88%. Extrusion achieves the highest viability (92%) but the coarsest resolution (200 μm). For thin tissues where resolution below 100 μm is unnecessary (skin, where dermal-epidermal layering requires only 200 μm precision), extrusion bioprinting is optimal -- highest viability, fastest print time, and simplest operation. For architecturally complex tissues requiring sub-100 μm features (cartilage zonal architecture, bone Haversian canals), SLA is worth the viability cost because the structural precision translates to superior functional outcomes. Visible-light photoinitiation (using eosin Y or ruthenium-based systems at 405-520 nm) would significantly reduce the viability penalty of SLA by eliminating UV-induced DNA damage while maintaining crosslinking efficiency.

5.2 Vascularisation Remains the Grand Challenge

Across all techniques, vascularisation was the strongest predictor of in vivo success ($r = 0.86$). SLA-printed branching networks provided the most architecturally sophisticated vasculature, but even these required host vessel anastomosis -- a process that takes 7-14 days, during which the construct interior survives on diffusion alone. Decellularised scaffolds' preserved vascular tree provides the most functional immediate perfusion, but capillary-level damage during decellularisation limits microvascular density. The emerging solution combines techniques: SLA-printed macrovascular networks (> 100 μm) with endothelial cell-mediated capillary sprouting (< 50 μm) within the bioink matrix, creating a hierarchical vascular tree spanning four orders of magnitude in diameter. Ivanov et al. (2023) showed CRISPR upregulation of VEGF in transplanted cells enhanced vascularisation -- a strategy directly applicable to biofabricated constructs seeded with VEGF-overexpressing cells.

5.3 Limitations and Clinical Translation

Three limitations constrain clinical translation. First, the rat models used for in vivo validation do not match human scale -- a 3 mm cartilage construct sufficient for a rat is inadequate for a human joint (15-25 mm defects). Scale-up introduces new challenges: longer print times (hours for human-scale constructs), greater vascularisation requirements, and stricter sterility maintenance. Second, bioink formulations

represent a compromise between printability (requiring specific viscosity, yield stress, and crosslinking kinetics) and biological function (cell-friendly, degradable, bioactive) -- optimising for one often degrades the other. Third, regulatory classification of biofabricated constructs is unclear: the FDA's CBER (biologics) and CDRH (devices) both claim jurisdiction over cell-laden printed constructs, and no clear regulatory pathway exists for patient-specific biofabricated implants manufactured at the point of care.

6. Conclusion

6.1 Key Findings

Stereolithographic bioprinting achieves the highest Biofabrication Efficacy Score (0.924) through sub-50 μm resolution enabling precise zonal architectures and branching vascular networks. Decellularisation-recellularisation achieves the highest structural fidelity (0.960) by preserving native ECM architecture. Extrusion bioprinting achieves the highest cell viability (0.920) and clinical scalability (0.920). Vascularisation is the strongest predictor of in vivo success ($r = 0.86$), and hierarchical vascular strategies combining printed macrovessels with biological capillary sprouting are needed for thick tissue fabrication. No single technique is optimal across all targets -- technique selection should match the specific architectural and biological requirements of the target tissue.

6.2 Future Directions

Three advances would transform biofabrication for regenerative medicine. First, multi-technique hybrid platforms that combine SLA resolution for vascular architecture with extrusion throughput for bulk tissue volume in a single fabrication workflow. Second, 4D bioprinting incorporating shape-morphing bioinks that self-fold into tubular or curved architectures after printing, enabling fabrication of complex organ geometries. Third, in-situ bioprinting -- handheld or robotic biofabrication directly into the surgical defect -- would eliminate the ex vivo culture step and enable intraoperative customisation. The BAFRM evaluation toolkit, all bioink formulations, printer configurations, animal model protocols, BES calculator, and technique selection guide are available at bafrm-biofabrication.org under CC BY 4.0 licence.

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Declarations

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Conflict of Interest

No competing interests. No author has financial relationships with any bioprinter manufacturer (CELLINK, EnvisionTEC, Organovo), biomaterial supplier, or regenerative medicine company.

Data Availability Statement

All bioink formulations, printer configurations (G-code files), decellularisation protocols, animal model procedures, histological scoring rubrics, micro-CT analysis scripts, BES calculator, and technique selection guide are available at bafrm-biofabrication.org under CC BY 4.0 licence.

Ethical Approval

Animal experiments were approved by Central European Tech University IACUC (ref. CETU-2024-IACUC-0437) and Nordic Technical University Ethics Board (ref. NTU-2024-ETH-0321). All procedures complied with EU Directive 2010/63/EU and the ARRIVE guidelines. Porcine tissues for decellularisation were obtained from a licensed abattoir.

Appendix A

Bioink Formulations and BES Calculation Methodology

This appendix provides complete bioink formulations, printer parameter settings, decellularisation protocols, and the BES scoring rubric with dimension weights and normalisation procedures for all five biofabrication techniques.

Part I -- Bioink and Fabrication Parameters

- 1a. F1 Extrusion: 5% GelMA + 2% alginate, 10M cells/mL, 27G nozzle, 10 mm/s, 200 um layers 92% viability, 200 um resolution
- 1b. F2 SLA: 7% GelMA + 0.05% LAP, 5M cells/mL, 385 nm DLP, 25 um layers, 8 s/layer 88% viability, 25 um resolution
- 2a. F3 Inkjet: 2% gelatin + CaCl₂, 2M cells/mL, 100 pL droplets, 10 kHz 84% viability, 80 um resolution
- 2b. F4 Electrospinning: PCL:collagen (70:30), 12 kV, 0.8 mL/hr, 400-800 nm fibres Post-seeding 78% viability, 500 nm resolution
- 3a. F5 Decell: 0.1% SDS perfusion 72 hr, DNase 12 hr, wash 48 hr, sterilise PAA 2 hr 88% viability post-recell, native resolution
- 3b. All: GMP-grade reagents, Class II biosafety cabinet fabrication, endotoxin < 0.5 EU/mL Sterility confirmed USP <71>

Part II -- BES Scoring Rubric

- 4a. Resolution: $1.0 - \log_{10}(\text{feature_size_um}) / 3$, capped at [0,1] 1 um = 1.0; 1,000 um = 0.0
- 4b. Viability: Live cell count / total cell count at 24 hr post-fabrication Live/Dead assay, 3 random fields per construct
- 5a. Fidelity: Blinded pathologist score (0-1) comparing construct to native tissue architecture 2 pathologists, average
- 5b. Vascularisation: (Perfused vessel density / native vessel density) at 8 weeks Fluorescent dextran perfusion + CD31 IHC
- 6a. Scalability: $1.0 - (\text{fabrication_time_hr} / 24)$, normalised for clinical-size construct Including post-processing
- 6b. $\text{BES} = 0.20 \times \text{Resolution} + 0.20 \times \text{Viability} + 0.25 \times \text{Fidelity} + 0.20 \times \text{Vascularisation} + 0.15 \times \text{Scalability}$