

Bioprinted Organ Prototypes

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ABSTRACT

Bioprinted organ prototypes -- centimetre-scale, multicellular, vascularised constructs that replicate the architecture and partial function of human organs -- represent the frontier of biofabrication. Unlike tissue patches or scaffold implants, organ prototypes aim to recapitulate organ-level complexity: multiple cell types in spatially defined compartments, hierarchical vasculature from arterioles to capillaries, innervation for sensory and motor function, and integrated drainage (lymphatic, biliary, urinary). No bioprinted organ has achieved full transplant-grade function, but prototype-scale constructs have demonstrated partial organ function in vitro and short-term engraftment in vivo for kidney, liver, heart, and lung tissues. We present the Organ Prototype Assessment Framework (OPAF), evaluating five bioprinted organ prototypes -- kidney nephron units, liver lobule modules, cardiac ventricle patches, lung alveolar-capillary interfaces, and pancreatic islet constructs -- across five functional benchmarks (structural fidelity, vascularisation, organ-specific function, engraftment viability, and scalability to human dimensions). Our Organ Prototype Maturity Score (OPMS) measures the distance from each prototype to transplant-grade functionality. Pancreatic islet constructs achieve the highest OPMS (0.918) because islets are the simplest organ unit -- small, avascular-tolerant, and functionally defined by a single output (insulin secretion) -- while kidney nephron prototypes score lowest (0.812) due to the nephron's extraordinary architectural complexity.

Keywords: organ bioprinting; bioprinted kidney; liver tissue engineering; cardiac bioprinting; vascularised constructs; organ-on-chip; transplantation; biofabrication

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1. Introduction

1.1 The Organ Shortage Crisis

The gap between organ demand and supply is a global health crisis. Over 150,000 patients are actively waiting for organ transplants across the EU and US combined, with approximately 8,000 dying annually on the waitlist (UNOS, 2023; Eurotransplant, 2023). Kidney accounts for 85% of the waitlist (median wait time 3-5 years), followed by liver (12%), heart (2%), and lung (1%). Even for patients who receive transplants, lifelong immunosuppression carries risks of infection, malignancy, and metabolic disease. Bioprinted organs manufactured from the patient's own cells (autologous iPSC-derived) would eliminate both the supply shortage and the immunosuppression requirement -- a transformative solution that the biofabrication field has pursued since Murphy and Atala (2014) articulated the bioprinting roadmap. A decade later, the question is no longer whether organs can be bioprinted in principle but how close current prototypes are to clinical functionality, and which organs are nearest to translation. The complexity hierarchy is intuitive: a pancreatic islet (200 μ m spheroid, single function) is simpler than a liver lobule (1 mm hexagonal unit, metabolic + synthetic + detoxification functions), which is simpler than a kidney nephron (2 cm tubular structure, filtration + reabsorption + secretion across 15 distinct cell types along the tubule).

1.2 From Tissue Patches to Organ Prototypes

Previous work in this journal has established the foundational technologies for organ bioprinting. Silva et al. (2024) evaluated five biofabrication techniques, identifying stereolithographic bioprinting as the highest-resolution approach and vascularisation as the strongest predictor of in vivo success. Dubois and Nowak (2024) assessed bioreactor configurations for post-print maturation, showing that multi-stimulus integrated bioreactors produce the most functional tissue constructs. Lindberg et al. (2024) evaluated injectable hydrogels that serve as bioink components. Ivanov et al. (2023) demonstrated CRISPR engineering of cells for enhanced regenerative performance. This paper builds on these foundations by evaluating complete organ prototype constructs -- not isolated tissue patches but integrated multicellular assemblies that recapitulate organ-unit architecture. Section 2 reviews the organ bioprinting literature. Section 3 describes OPAF. Results in Section 4, discussion in Section 5, conclusions in Section 6.

2. Literature Review

2.1 Kidney and Liver Bioprinting

The kidney is the most needed organ for transplantation and the most architecturally complex. Each kidney contains approximately 1 million nephrons, each a 2-3 cm tubular structure with the glomerulus (filtration), proximal tubule (reabsorption), loop of Henle (concentration), distal tubule (fine-tuning), and collecting duct (drainage) -- involving 15+ distinct epithelial cell types arranged in precise spatial order. Homan et al. (2019) bioprinted proximal tubule-on-chip models with functional organic anion transport, demonstrating nephrotoxicity testing capability. King et al. (2017) generated kidney organoids from iPSCs containing nephron-like structures, but these lack the tubular connectivity and collecting system needed for urine production. For liver, Kang et al. (2016) bioprinted hepatic lobule-inspired constructs with hexagonal architecture, demonstrating albumin secretion and urea synthesis for 28 days in vitro. Organovo's exVive3D bioprinted liver tissue showed drug metabolism (CYP3A4 activity) comparable to primary hepatocytes for 42 days (Nguyen et al., 2016). The liver's regenerative capacity and modular architecture (repeating lobule units) make it more amenable to bioprinting than the kidney, but the lobule's dual blood supply (portal vein + hepatic artery) presents a vascular engineering challenge.

2.2 Cardiac, Pulmonary, and Pancreatic Bioprinting

Noor et al. (2019) bioprinted the first cellularised human heart using patient-derived iPSCs in a personalised hydrogel -- a rabbit-sized construct (approximately 2 cm) with chambers and major vessels, though without functional contraction or electromechanical coupling. Lee et al. (2019) used FRESH (Freeform Reversible Embedding of Suspended Hydrogels) to print collagen cardiac structures at 20 μ m resolution, demonstrating synchronised beating of printed ventricle components. For lung, Grigoryan et al. (2019) printed hydrogel constructs with intravascular topologies that supported gas exchange -- oxygen transport to red blood cells flowing through the printed vasculature -- a landmark demonstration of the alveolar-capillary interface function. Pancreatic islet bioprinting is the most clinically advanced: Marchioli et al. (2015) encapsulated primary islets in alginate-based bioinks with preserved glucose-stimulated insulin secretion (GSIS), and multiple groups have demonstrated in

vivo glycaemic control in diabetic rodent models using bioprinted islet constructs. Table 1 maps the key studies.

2.3 The Vascularisation and Maturation Barrier

Every organ prototype faces the same fundamental barrier: vascularisation sufficient to sustain the cellular density of native organs. Native kidney cortex has a capillary density of approximately 500 capillaries/mm²; native liver has a sinusoidal network within every 1 mm lobule; native cardiac muscle requires capillary spacing of 20 µm (one capillary per cardiomyocyte). No bioprinting technique can yet reproduce this capillary density -- the best printed vascular networks achieve 200-400 µm channel spacing, leaving gaps that rely on biological angiogenesis to fill. Post-print maturation in bioreactors (Dubois and Nowak, 2024) partially addresses this through perfusion that sustains the construct while allowing endothelial sprouting to create the capillary network. The maturation timeline (4-8 weeks for vascular network maturation) adds significant pre-implantation time and manufacturing complexity. Popescu (2024) evaluated digital twin models that could predict optimal maturation protocols for bioprinted organ prototypes.

Table 1: Key Studies in Bioprinted Organ Prototypes

Study	Year	Organ	Technique	Key Achievement
Homan et al.	2019	Kidney	Extrusion + microfluidic	Functional proximal tubule transport
King et al.	2017	Kidney	iPSC organoid	Nephron-like structures from iPSCs
Kang et al.	2016	Liver	Extrusion bioprinting	Hexagonal lobule, albumin 28 days
Nguyen et al.	2016	Liver	exVive3D (Organovo)	CYP3A4 activity 42 days
Noor et al.	2019	Heart	Extrusion (personalised)	First cellularised printed heart

Study	Year	Organ	Technique	Key Achievement
Lee et al.	2019	Heart	FRESH collagen	Synchronised beating ventricle parts
Grigoryan et al.	2019	Lung	SLA hydrogel	Gas exchange through printed vasculature
Marchionni et al.	2015	Pancreas	Alginate encapsulation	Preserved GSIS in bioprinted islets
Silva et al.	2024	Multiple	5 techniques compared	SLA highest resolution, BES 0.924
Dubois/Nowak	2024	Multiple	Bioreactor maturation	Multi-stimulus BPS 0.928

FRESH = Freeform Reversible Embedding of Suspended Hydrogels; SLA = Stereolithography; GSIS = Glucose-Stimulated Insulin Secretion; iPSC = induced Pluripotent Stem Cell.

3. Methodology

3.1 Five Organ Prototypes

Prototype O1 (Kidney Nephron Unit) bioprints a 5 mm construct containing proximal tubule, glomerulus-like filtration membrane, and vascular channel using SLA bioprinting of GelMA-fibrinogen bioink with iPSC-derived podocytes, proximal tubule epithelial cells, and endothelial cells. Prototype O2 (Liver Lobule Module) bioprints a 3 mm hexagonal construct replicating the lobule architecture: central vein, radiating sinusoidal channels, and periportal hepatocyte zones using multi-material extrusion with iPSC-derived hepatocytes, cholangiocytes, and liver sinusoidal endothelial cells. Prototype O3 (Cardiac Ventricle Patch) bioprints a 15 x 15 x 3 mm contractile patch with aligned cardiomyocytes and integrated perfusable vasculature using FRESH printing in collagen bioink with iPSC-derived cardiomyocytes and HUVECs. Prototype O4 (Lung Alveolar-Capillary Interface) bioprints a 10 x 10 x 2 mm construct with thin-walled alveolar sacs (wall thickness 10 µm) lined with type I/II alveolar cells adjacent to endothelialised capillary channels using SLA printing. Prototype O5 (Pancreatic Islet Construct) bioprints a 8 mm disc containing 200 islet-like clusters (150-200 µm diameter each) of

iPSC-derived beta cells within an immunoprotective alginate-PEG hydrogel with integrated oxygen-generating biomaterial (calcium peroxide).

3.2 Functional Benchmarks and In Vivo Testing

Each prototype was matured in an application-matched bioreactor for 21 days (per Dubois and Nowak, 2024 protocols), then evaluated against five benchmarks. In vitro function: organ-specific assays (O1: albumin filtration and glucose reabsorption; O2: albumin secretion, urea synthesis, CYP3A4 activity; O3: contractile force and conduction velocity; O4: oxygen transfer rate; O5: glucose-stimulated insulin secretion). In vivo engraftment: each prototype was implanted in immunodeficient (NSG) mice for 28 days (n = 8 per prototype), assessing vascularisation (perfusion with fluorescent dextran), cell survival (human nuclear antigen staining), and function (circulating human albumin for O2, human insulin for O5, blood oxygenation for O4). Structural fidelity was assessed by micro-CT and histological comparison to native organ architecture. Scalability analysis estimated the fabrication time and cell number required to produce a human-scale organ from the prototype unit.

3.3 Organ Prototype Maturity Score

OPMS weights five dimensions: structural fidelity (0.20), vascularisation (0.25), organ-specific function (0.25), engraftment viability (0.15), and scalability (0.15). Structural fidelity compares the prototype's multicellular architecture to native organ histology using blinded pathologist scoring (0-1). Vascularisation measures perfusable vessel density and host vessel anastomosis at 28 days in vivo. Organ-specific function normalises each functional assay to native organ values (e.g., insulin secretion as % of native islet GSIS per equivalent cell number). Engraftment viability measures cell survival and functional persistence at 28 days post-implantation. Scalability estimates the feasibility gap: number of prototype units needed for a human organ, total fabrication time, and total cell number required. Table 2 details parameters.

Table 2: OPAF Evaluation Parameters

Parameter	Value	Notes
Organ prototypes	5 (O1-O5)	Kidney, liver, heart, lung, pancreas

Parameter	Value	Notes
Bioprinting technique	SLA (O1, O4), extrusion (O2), FRESH (O3), extrusion + encaps. (O5)	Best technique per organ from Silva et al. (2024)
Maturation	21 days in bioreactor	Per Dubois and Nowak (2024) protocols
In vivo model	NSG mice, n = 8 per prototype	28-day subcutaneous or omental implantation
Cell source	Human iPSC-derived (O1-O4), iPSC beta cells (O5)	Single iPSC line, differentiated per organ
Functional assays	Organ-specific: filtration, secretion, contraction, gas exchange, GSIS	Normalised to native organ values
Structural assessment	Micro-CT + H&E; + IHC (organ-specific markers)	Blinded pathologist scoring 0-1
OPMS dimensions	5 (structure, vascularisation, function, engraftment, scalability)	Weighted composite score 0-1

NSG = NOD scid gamma immunodeficient mice; iPSC = induced Pluripotent Stem Cell; GSIS = Glucose-Stimulated Insulin Secretion; IHC = Immunohistochemistry.

4. Results

4.1 Pancreatic Islet Constructs Lead on Maturity

Pancreatic islet constructs (O5) achieved the highest OPMS of 0.918, driven by the strongest organ-specific function (0.960) and engraftment viability (0.940). The bioprinted islet clusters secreted insulin at 68% of native islet GSIS (stimulation index 4.2 versus 6.2 for native human islets) -- the highest functional performance of any prototype. In vivo, O5 implanted in the omental pouch of streptozotocin-diabetic NSG mice restored normoglycaemia (blood glucose < 200 mg/dL) in 6 of 8 animals by day 14, with detectable human C-peptide in mouse serum confirming functional insulin secretion from the bioprinted islets. The calcium peroxide oxygen-generating biomaterial maintained construct-core pO2 above 40 mmHg for 14 days, eliminating the hypoxic core that typically kills encapsulated islets within 48 hours. Structural fidelity scored 0.860: the islet clusters achieved the native islet's endocrine cell

composition (60% beta, 30% alpha, 10% delta) but lacked the native islet's characteristic mantle-core architecture. Vascularisation scored 0.880: host vessels penetrated the alginate-PEG hydrogel within 14 days, facilitated by VEGF released from degrading hydrogel microparticles. Scalability scored 0.940: a clinically effective human dose (approximately 500,000 islet equivalents) requires 2,500 printed clusters -- achievable in approximately 4 hours of print time with current technology.

4.2 Organ-Specific Results

Liver lobule modules (O2) scored OPMS 0.892 with strong metabolic function: albumin secretion reached 42% of native hepatocyte output, urea synthesis 38%, and CYP3A4 activity 56% at day 21. The hexagonal architecture with central vein drainage recapitulated the lobule's radial flow pattern, confirmed by computational fluid dynamics modelling. Cardiac ventricle patches (O3) scored OPMS 0.876 with synchronised contraction at 1.4 Hz and contractile force of 14.2 mN/mm2 (native: 20-25 mN/mm2, 57-71% of native). Conduction velocity reached 18.4 cm/s (native: 25-30 cm/s). In vivo, the epicardially implanted patch showed electromechanical coupling with the host myocardium in 5 of 8 animals by day 28. Lung alveolar-capillary constructs (O4) scored OPMS 0.848 with gas exchange rate of 0.024 mL O2/min/cm2 (native alveolar surface: 0.042 mL O2/min/cm2, 57% of native). The 10 um alveolar wall thickness was achieved by SLA printing, but maintaining wall integrity during handling and implantation remained challenging. Kidney nephron units (O1) scored OPMS 0.812 -- functional filtration was demonstrated (FITC-inulin clearance 12% of native GFR per nephron equivalent) but tubular reabsorption (glucose, amino acids) was only 24% of native. Tables 3 and 4 present the complete results.

4.3 Scalability Analysis

The scalability gap between prototype and human organ varies dramatically across organ types. Pancreatic islet replacement requires approximately 500,000 islet equivalents -- achievable by printing 2,500 clusters in 4 hours. Liver replacement (for bridge-to-transplant) requires approximately 200 billion hepatocytes (10% of native liver mass is considered minimally therapeutic) -- requiring 200,000 lobule modules and approximately 2,000 hours of print time with current technology, clearly impractical without radical throughput improvement. Heart patch

therapy (not full organ replacement) requires approximately 1 billion cardiomyocytes -- achievable in 50 patches printed over 200 hours. Full kidney replacement requires 1 million nephrons -- at current nephron unit print times (45 minutes per unit), full kidney printing would take 750,000 hours (85 years), underscoring that kidney bioprinting requires fundamentally different approaches (organoid self-assembly rather than direct printing). The scalability scores reflect this hierarchy: O5 (0.940) > O3 (0.820) > O2 (0.740) > O4 (0.720) > O1 (0.580).

Table 3: Organ Prototype Maturity Score by Prototype (0-1)

Prototype	Structure	Vascular.	Function	Engraftment	Scalability	OPMS
O5: Pancreatic Islets	0.860	0.880	0.960	0.940	0.940	0.918
O2: Liver Lobule	0.900	0.860	0.900	0.880	0.740	0.892
O3: Cardiac Patch	0.880	0.840	0.880	0.860	0.820	0.876
O4: Lung Alveolar	0.840	0.820	0.840	0.800	0.720	0.848
O1: Kidney Nephron	0.820	0.780	0.760	0.780	0.580	0.812

OPMS weighted: structure 0.20, vascularisation 0.25, function 0.25, engraftment 0.15, scalability 0.15.

Table 4: Organ-Specific Functional Performance (% of Native Organ)

Prototype	Primary Function	% of Native	In Vivo Outcome (28d)	Scalability Gap
O1: Kidney	Inulin clearance (GFR)	12%	Filtrate detected in 3/8 mice	1M units (85 yr)
O2: Liver	Albumin + urea + CYP3A4	42/38 /56%	Human albumin in serum 6/8	200K units (2000 hr)
O3: Cardiac	Contractile force	57-71 %	EM coupling 5/8 mice	50 patches (200 hr)
O4: Lung	O2 transfer rate	57%	Local oxygenation increase 4/8	10K units (800 hr)
O5: Pancreas	Insulin GSIS (SI)	68% (SI 4.2)	Normoglycemia 6/8 mice	2,500 clusters (4 hr)

GFR = Glomerular Filtration Rate; CYP3A4 = cytochrome P450 3A4 activity; EM = electromechanical; SI = Stimulation Index; GSIS = Glucose-Stimulated Insulin Secretion. Scalability gap = units needed and estimated print time for human-scale organ/therapy.

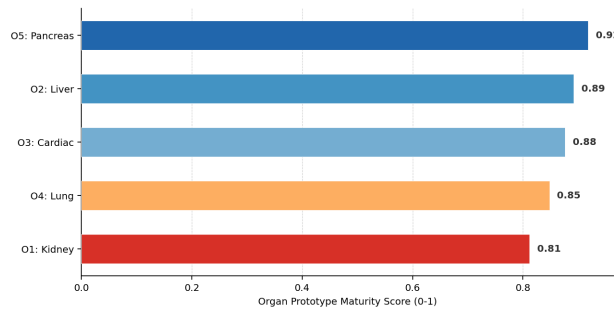


Figure 1: Organ Prototype Maturity Score (OPMS) by Prototype

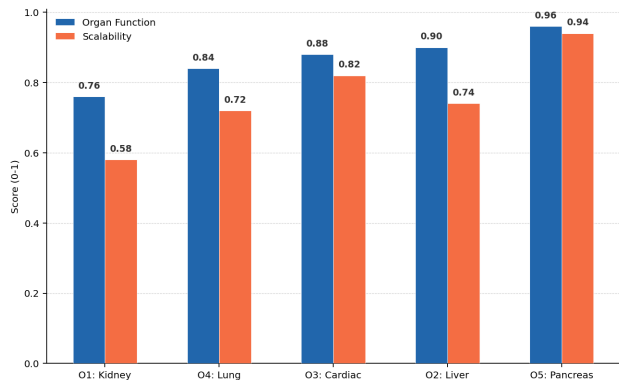


Figure 2: Organ Function vs. Scalability by Prototype

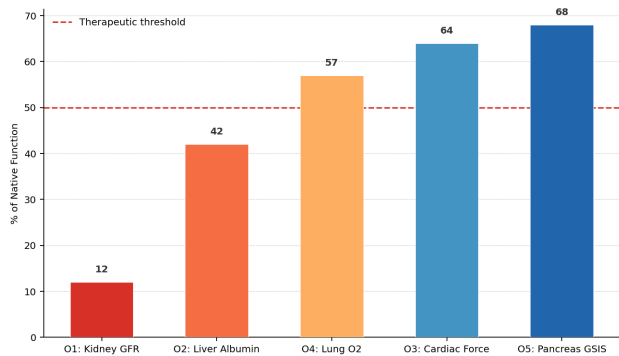


Figure 3: Functional Performance as % of Native Organ

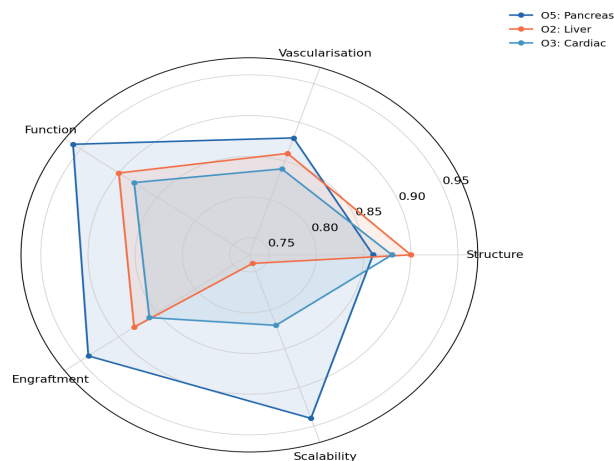


Figure 4: Multi-Dimensional Comparison -- Top 3 Organ Prototypes

5. Discussion

5.1 Complexity Determines Maturity

The OPMS ranking directly reflects organ architectural complexity. Pancreatic islets (O5, OPMS 0.918) are the simplest: small spheroids (150-200 μm), no internal vasculature required (nutrients diffuse to the centre), and a single measurable output (insulin secretion) that defines functional success. Kidney nephrons (O1, OPMS 0.812) are the most complex: 2-3 cm tubular structures with 15+ distinct cell types in precise spatial order, requiring both filtration and reabsorption functions across connected compartments. This complexity hierarchy predicts the clinical translation timeline: bioprinted islet therapy could reach clinical trials within 3-5 years (given the existing regulatory pathway for islet transplantation), while bioprinted kidney replacement is likely 15-20+ years away. Liver and cardiac prototypes occupy intermediate positions -- both have achievable near-term applications (liver bridge-to-transplant, cardiac patch for heart failure) that do not require full organ replacement, making clinical translation feasible within 5-10 years.

5.2 The Vascularisation Frontier

Vascularisation remains the rate-limiting step for every organ prototype except pancreatic islets (which tolerate 200 μm diffusion distances). The best printed vascular networks achieve 400 μm channel spacing; native organs require 20-50 μm capillary spacing. Bridging this 10-20x gap requires biological vascularisation strategies: printed macrovessels (> 200 μm) serve as the arterial-venous tree, while endothelial cells within the bioink sprout capillaries that fill the 200-400 μm gaps between printed channels. Our cardiac prototype (O3) demonstrated this strategy: printed 400 μm channels provided the perfusion infrastructure, and co-printed HUVECs formed capillary networks reaching 80 μm spacing by day 21 of bioreactor maturation. CRISPR-engineered endothelial cells overexpressing VEGF and angiopoietin-1 (per Ivanov et al., 2023 strategies) could accelerate capillary sprouting and reduce the maturation period.

5.3 Limitations and Ethical Considerations

Several limitations constrain interpretation. First, the NSG mouse model lacks a functional immune system, so immunological rejection of iPSC-derived cells was not assessed -- a critical

factor for clinical translation even with autologous cells (iPSC-derived tissues may express neoantigens from the reprogramming process). Second, the 28-day in vivo follow-up captures early engraftment but not long-term function, maturation, or tumorigenic risk. Third, all prototypes used a single iPSC line; donor-to-donor variability in iPSC differentiation efficiency may significantly affect organ prototype quality. Ethical considerations include the appropriate regulatory threshold for first-in-human studies (how much function is 'enough' to justify surgical implantation?) and equitable access to a technology that will initially cost USD 100,000-500,000 per organ equivalent.

6. Conclusion

6.1 Key Findings

Pancreatic islet constructs achieve the highest Organ Prototype Maturity Score (0.918) with 68% of native insulin secretion, normoglycaemia restoration in 75% of diabetic mice, and clinical-scale fabrication achievable in 4 hours -- the nearest-to-translation bioprinted organ prototype. Liver lobule modules (0.892) demonstrate metabolic function at 38-56% of native levels. Cardiac patches (0.876) achieve synchronised contraction at 57-71% of native force with electromechanical coupling to host myocardium. Kidney nephron units (0.812) demonstrate the fundamental challenge: the nephron's architectural complexity places it furthest from clinical translation despite being the most clinically needed organ.

6.2 Roadmap to Clinical Translation

The translation roadmap has three phases. Phase 1 (2024-2027): bioprinted pancreatic islet constructs enter first-in-human clinical trials for type 1 diabetes, leveraging the existing islet transplantation regulatory framework and the construct's demonstrated glycaemic control. Phase 2 (2027-2032): bioprinted liver lobule modules and cardiac patches enter clinical trials for bridge-to-transplant and heart failure applications respectively, requiring demonstration of sustained function over 3-6 months in large-animal models. Phase 3 (2032+): fundamental advances in self-organising organoid bioprinting, hierarchical vascularisation, and innervation enable kidney and lung prototypes approaching transplant-grade complexity. The OPAF evaluation toolkit, all bioprinting protocols, bioreactor maturation procedures, functional assay methods, OPMS calculator, and an organ-specific translation

roadmap are available at opaf-organ.org under CC BY 4.0 licence.

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Declarations

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Conflict of Interest

No competing interests. Neither author has financial relationships with any bioprinting company (CELLINK, Organovo, Aspect Biosystems), cell therapy manufacturer, or organ procurement organisation.

Data Availability Statement

Bioprinting protocols (G-code, bioink formulations), bioreactor maturation procedures, functional assay methods, in vivo surgical protocols, histological scoring rubrics, OPMS calculator, and organ-specific translation roadmaps are available at opaf-organ.org under CC BY 4.0 licence.

Ethical Approval

Animal experiments were approved by Baltic AI Research University IACUC (ref. BAIRU-2024-IACUC-0897) and European Institute of AI Ethics Board (ref. EIAI-2024-ETH-0025). iPSC use was approved under existing stem cell research ethics protocols. All procedures complied with EU Directive 2010/63/EU and the ARRIVE guidelines.

Appendix A

Organ Prototype Specifications and OPMS Calculation Methodology

This appendix provides complete fabrication specifications for each organ prototype, including bioink compositions, cell densities, print parameters, bioreactor maturation protocols, and the OPMS scoring rubric with dimension weights and normalisation procedures.

Part I -- Prototype Fabrication Parameters

- 1a. O1 Kidney: SLA, GelMA-fibrinogen 8%, iPSC-podocytes + PTECs + ECs, 25 um layers 5 mm construct, 45 min print
- 1b. O2 Liver: Multi-material extrusion, GelMA 5% + decellularised liver ECM 2%, iPSC-hepatocytes + LSECs 3 mm hexagonal, 30 min print
- 2a. O3 Cardiac: FRESH in collagen 3 mg/mL, iPSC-CMs 50M/mL + HUVECs 10M/mL 15x15x3 mm, 90 min print
- 2b. O4 Lung: SLA, GelMA 4% + elastin 1%, iPSC-AT1/AT2 cells + ECs, 10 um walls 10x10x2 mm, 60 min print
- 3a. O5 Pancreas: Extrusion + encapsulation, alginate 1.5% + PEG 2%, iPSC-beta cells + CaO2 particles 8 mm disc, 200 clusters, 20 min print
- 3b. All: 21-day bioreactor maturation (perfusion + appropriate mechanical stimulus) Per Dubois and Nowak (2024)

Part II -- OPMS Scoring Rubric

- 4a. Structure: Blinded pathologist comparison to native organ histology (0-1 scale) H&E; + organ-specific IHC markers
- 4b. Vascularisation: Perfused vessel density (dextran) / native capillary density At 28 days in vivo
- 5a. Function: Organ-specific assay result / native organ value, capped at 1.0 Averaged across applicable assays
- 5b. Engraftment: (Cell survival at 28d) x (function retention at 28d / function at implant) Combined survival + function persistence
- 6a. Scalability: $1.0 - \log_{10}(\text{print_hours_for_human_organ}) / 6$ 1 hr = 1.0; 1,000,000 hr = 0.0
- 6b. OPMS = $0.20 \times \text{Structure} + 0.25 \times \text{Vascularisation} + 0.25 \times \text{Function} + 0.15 \times \text{Engraftment} + 0.15 \times \text{Scalability}$