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Personalized TPMS scaffolds for maxillary bone regeneration: Inverse design, simulation, and stereolithographic fabrication

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Abstract

This paper describes a method of designing and producing patient-specific scaffold prototypes for bone tissue regeneration by means of reverse engineering techniques and triply periodic minimum surface (TPMS) structures to create a tissue scaffold-type implant. A 3-D bone defect model was reconstructed from DICOM tomographic images of an actual clinical case. Five scaffold variants were designed on Meshmixer and nTop™ computer-aided design (CAD) software, each variant with a different TPMS topology (Gyroid, Schwarz P, Diamond, Lidinoid, and Split P), while maintaining a constant cell size of 1 mm and a wall thickness of 0.16 mm. The scaffolds' behaviour under a compressive load of 698 N was computationally evaluated using finite element analysis (FEA) in which the Lidinoid and Schwarz P structures were found to exhibit the least deformation. The scaffolds were fabricated by stereolithography (SLA) and a morphological analysis was performed in which interconnected porosity and pore dimensions of between 100 μm and 500 μm were identified, conditions considered optimal for cell infiltration and vascularization. These findings demonstrate the feasibility of a proof-of-concept workflow for designing and fabricating patient-specific TPMS scaffold prototypes and identify topology-dependent differences in their predicted mechanical response and visible pore morphology. Further studies using clinically suitable osteoconductive and bioresorbable materials are required before considering implantation.

Keywords: Triply periodic minimum surface; Tissue engineering; Bone regeneration; Additive manufacturing; Finite element analysis; Photopolymer resin

1. Introduction

Tissue engineering is a rapidly growing multidisciplinary field offering promising solutions for the regeneration and repair of damaged tissues and organs¹. As bone defects caused by trauma or disease pose significant clinical challenges that affect the patient's quality of life, the design and manufacture of tissue scaffolds represent a substantial advance in tissue engineering². Current therapies using autologous and allogeneic grafts often face challenges related to tissue availability, donor site morbidity and a risk of rejection or the transmission of disease. Given these limitations, the design and fabrication of customized biomimetic tissue scaffolds are emerging as a superior alternative, capable of guiding *in situ* bone regeneration and improving the long-term biological integration and functionality of the implant. These scaffolds not only adapt to the different morphologies of individual patients but also enhance biological interaction and functionality^{3,4}.

Tissue scaffolds are essential in tissue engineering as they provide mechanical support and a microenvironment that promotes cell growth, differentiation and migration for tissue regeneration^{1,5,6}. An effective scaffold must be: osteoconductive; biocompatible, with the capacity for direct tissue integration and absence of fibrous encapsulation; bioabsorbable, replaceable by native tissue; highly porous, to ensure the diffusion of nutrients and cell growth; and must also have a resistance similar to the tissue it will replace^{1,6-10}.

These characteristics represent general requirements for clinically implantable scaffolds; however, the present proof-of-concept study evaluates only the design, predicted mechanical response, manufacturability, and visible pore morphology of the fabricated prototypes.

To achieve this functional mimicry, it is important to consider the microarchitecture of native bone tissue, especially in regions like the maxilla, with a predominance of trabecular bone. This tissue is characterized by a highly porous structure, with values ranging from 50% to 90%, organized in a network of trabeculae that dictates both the biological response and the load distribution^{11,12}. In mechanical terms, this porous configuration results in a compressive strength that typically ranges between 1.5 and 9.3 MPa¹¹.

The internal architecture of the scaffold, especially its porosity and connectivity, plays a critical role in its biological and mechanical performance. Scaffold design seeks to mimic the natural extracellular matrix, balancing mechanical and biological properties for the regeneration of various tissues¹². Precise control of the internal geometry is fundamental, including the size and distribution of the pores, since pores larger than 100 μm are essential to allow vascularization and

fluid flow, although excessively large diameters can compromise the structural stability of the graft¹³. Optimizing a scaffold in this context thus involves not only using biocompatible materials but also achieving a volumetric and geometric balance. This is accomplished by using high-precision medical imaging to ensure that the customized design mimics the natural architecture and guarantees the implant's biological viability¹⁴⁻¹⁶.

Additive manufacturing (AM), especially stereolithography (SLA), has revolutionized the field in terms of manufacturing methods by enabling the layer-by-layer construction of complex structures with high dimensional accuracy and exceptional control over the porous geometry¹⁴⁻¹⁷. Among the various architectures that can be fabricated by AM, TPMS structures have gained significant attention in tissue engineering due to their unique properties^{18,19}. TPMS are mathematical surfaces with a zero mean curvature that exhibit a periodic crystalline structure^{2,20}, offering homogeneous stress distribution, improving compressive strength²¹⁻²³ and with highly interconnected porosity, which favours cell adhesion, migration and proliferation^{2,7,24-28}, and optimizes mass transport^{2,23,25,26, 28-30}, thus making them an ideal option for biomedical applications.

Experimental studies using vat-photopolymerized polymeric TPMS structures have demonstrated that scaffold morphology, printing fidelity, and mechanical response depend on the interaction between topology, relative density, wall thickness, and resin properties. Gabrieli et al.³¹ fabricated Gyroid and Schwarz P scaffolds by stereolithography using a commercial dental resin and experimentally characterized their morphology and compressive behaviour, showing that changes in wall thickness and relative density significantly influenced elastic modulus and compressive strength. Peloquin et al.³² similarly demonstrated that the volumetric accuracy, printability, and mechanical performance of photopolymer lattice structures depend on the lattice geometry, resin properties, and designed porosity. In addition, a previous study by the present authors experimentally characterized the tensile and compressive behaviour of Gyroid, Diamond, Schwarz P, Lidinoid, and Split-P structures fabricated using a dental photopolymer resin and comparable SLA processing conditions³³. These studies provide an experimental basis for selecting and interpreting photopolymer TPMS architectures, while also highlighting the importance of topology-specific manufacturing and mechanical behaviour.

The mechanical properties of TPMS can be adapted to those of human bone by adjusting parameters such as cell size and type or wall thickness^{18,24}. This customizing of the structures is fundamental to ensure a perfect fit at the injury site and optimizing osseointegration. From a

technical perspective, reverse engineering, using medical imaging procedures such as computed tomography (CT), can create precise 3D models of the patient's anatomy, focusing on the bone defect³⁴. These models serve as a basis for designing custom scaffolding that faithfully replicates the required geometry, guaranteeing an optimal and functional fit^{14,34}. This process combines medical image capture, 3D reconstruction and implant modelling, ensuring a perfect fit with the patient's anatomy¹⁴.

The tensile and compressive behaviour of the five TPMS families considered in the present study, Gyroid, Diamond, Schwarz P, Lidinoid, and Split-P, has previously been experimentally characterized using stereolithography-printed specimens manufactured from the same type of dental photopolymer resin and under comparable processing conditions³³. That previous study established topology- and density-dependent differences in stiffness, strength, and deformation mechanisms using standardized tensile and compression specimens.

Therefore, the purpose of the present work was not to repeat the mechanical characterization of standardized TPMS specimens, but to investigate how these architectures can be transferred to a patient-specific maxillary defect through a reverse-engineering, implicit-design, additive-manufacturing, and morphological-assessment workflow. Finite element analysis was incorporated as a supplementary comparative step to examine the predicted response of the patient-specific designs under a common simplified loading condition.

Accordingly, the present study should be regarded as a proof-of-concept investigation of the patient-specific design and fabrication workflow. The resin-printed structures were not intended for clinical implantation, and the study did not evaluate bioresorption, osteoconductivity, or long-term biological performance.

Despite the growing interest in TPMS scaffolds for bone engineering, the gap between the laboratory and the clinic still persists, aggravated by the lack of tangible medical applications and rigorous post-fabrication morphological analysis. This study therefore analyses TPMS structures in a tissue scaffold obtained by reverse engineering from medical images and evaluated using finite element analysis to assess their structural performance.

2. Materials and Methods

2.1. Materials

Phrozen Dental Model Water-Washable Resin was used as a prototyping material because of its compatibility with stereolithography and its ability to reproduce complex, thin-walled geometries. This rigid photopolymer was not considered a bioresorbable, osteoconductive, or permanently implantable material in the present study. Accordingly, the fabricated structures were used exclusively as proof-of-concept scaffold prototypes to evaluate the proposed patient-specific design, simulation, manufacturing, and morphological assessment workflow. The scaffold prototypes were manufactured using the AnyCubic Photon M3 Plus printer and the Anycubic Wash & Cure 3 Plus curing oven.

The mechanical response of stereolithography-printed Gyroid, Diamond, Schwarz P, Lidinoid, and Split-P structures manufactured using this resin and comparable printing conditions was experimentally characterized in a previous study by the authors³³. That study included standardized uniaxial tensile and compression tests across several wall-thickness levels. The present work builds upon that experimental foundation but focuses on the transfer of these TPMS architectures to a patient-specific anatomical geometry rather than on repeating coupon-level mechanical characterization.

The implants were designed on the Blue Sky Plan 4.13® medical software and Meshmixer 3.5 and nTop™ 5.28.4 CAD software. CAE Ansys™ (2025R2) software was used for the numerical simulation.

The morphological analysis was performed using a Stemi 508 stereomicroscope, Zeiss AxioCam 20838 camera and Zeiss Labscope software.

2.2. Methods

Figure 1 shows the method used in the development of the design and simulation of the tissue scaffold by means of flowcharts.

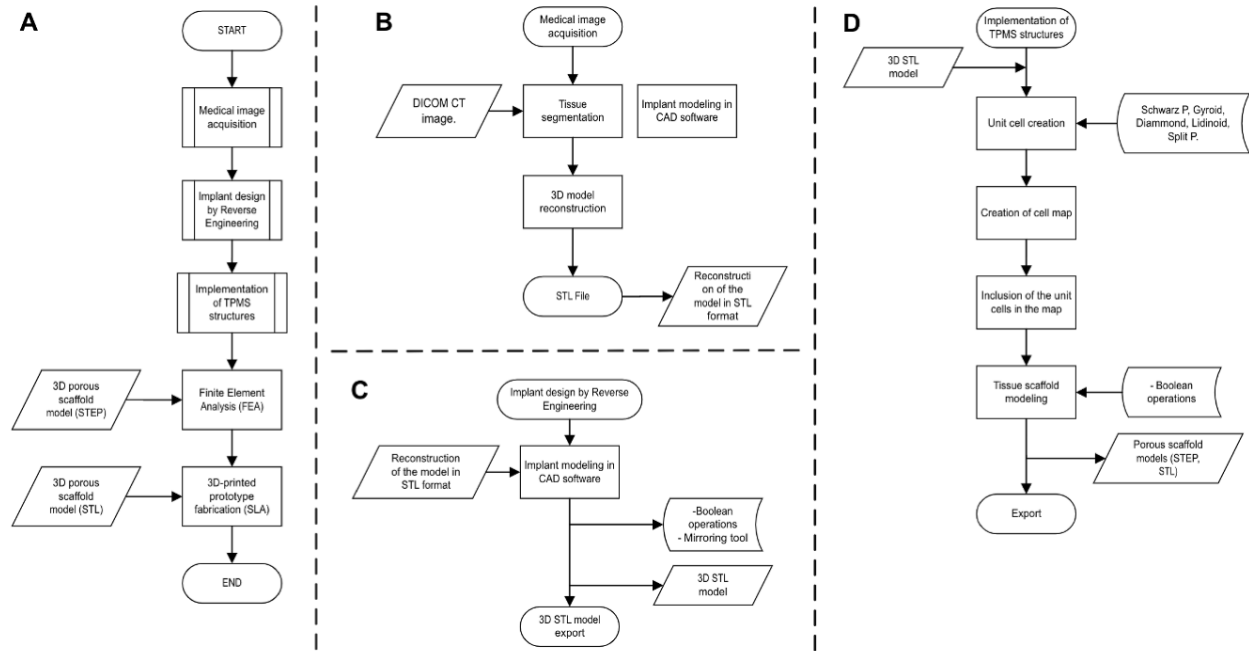


Figure 1. Workflow diagrams: A) General workflow of the design and manufacturing process, B) Acquisition and reconstruction of medical images, C) Implant design by reverse engineering, D) Implementation of TPMS structures.

2.2.1. Clinical case and CT image acquisition

Anonymized computed tomography images from an adult patient presenting bone resorption in the left posterior maxilla and absence of the first and second molars were used as the anatomical basis for the patient-specific scaffold design. The clinical images showed insufficient bone volume for the planned placement of two dental implants, indicating the need for prior reconstruction of the affected maxillary region.

The CT images were acquired using a Siemens Somatom Sensation 64 scanner under a standardized clinical imaging protocol. The tube voltage was set to 120 kV, the tube current to 150 mA, and the slice thickness to 0.6 mm. The images were exported in Digital Imaging and Communications in Medicine (DICOM) format for subsequent segmentation and three-dimensional reconstruction

These parameters were defined in accordance with the scanner's technical specifications and recent scientific reviews on dose optimization in dental tomography³⁵⁻³⁷.

2.2.2. Reverse Engineering

The DICOM images were imported into Blue Sky Plan® software. The bone tissue was then segmented, separating it from the surrounding soft tissues. 3D reconstruction (or slicing) was then

performed, a process that converts the segmented 2D sections into a digital 3D model. This operation generated a surface mesh in STL format. The desired implant model was finally obtained using Boolean operations and symmetry constraints on Meshmixer software.

2.2.3. TPMS Implementation

The implant model generated by Meshmixer was imported as a mesh into the nTop™ software for the implementation of TPMS structures. The five unit cell types defined were: Gyroid, Schwarz P, Diamond, Lidinoid, and Split P. These five TPMS topologies were selected for their interconnected pores, essential for tissue regeneration. Cell size values of 1 mm and a wall thickness of 0.16 mm generate pores of between 100 µm and 500 µm, an optimal range for promoting bone growth³⁸. These parameters facilitate the fabrication of the structures using SLA and allow a uniform comparison of the characteristics of the different TPMS.

2.2.4. Mechanical evaluation

The finite element assessment was included as a supplementary comparative stage of the proposed patient-specific design workflow. Its purpose was to compare the predicted response of the five scaffold configurations under identical simplified loading and boundary conditions. It was not intended to replace experimental testing of the final patient-specific scaffold or to demonstrate clinical load-bearing capacity.

A finite element analysis (FEA) was performed to evaluate the compressive mechanical behaviour of the different designed TPMS topologies. The digital models of the scaffolding were exported in STL format and imported into Ansys™ for numerical processing. Due to the geometric complexity of the structures the analysis was carried out from an equivalent voxelized representation. Spatial discretization was performed on a quadratic tetrahedral mesh, with a minimum element size of 0.2 mm to ensure adequate resolution in the areas of greatest mechanical interest.

A uniform compressive load of 698 N was applied, equivalent to the maximum functional masticatory force (MFMF) recorded in adults³⁹, this load case was adopted as a standardized reference condition to enable a consistent comparative assessment of the five TPMS configurations. The boundary conditions were defined with complete fixation at the base of the scaffolding, simulating rigid anchorage to the bone interface. An equivalent modulus of elasticity of 1542 MPa and an equivalent density of 1160 kg/m³ were used for the Phrozen Water-Washable Dental Model Resin material⁴⁰, using Formulas I and II. All the simulation parameters are defined in **Table 1**.

Table 1. Parameters used in the FEA analysis.

Parameter	Specification
Simulation software	Ansys 2025 R2
Material used	Phrozen Water-Washable Dental Model Resin
Elasticity Modulus	1542MPa
Density	1160 kg/m ³
Geometric Order	Quadratic
Minimum size of element	0.2 mm
Loading conditions	698 N Axial compression
Boundary conditions	Complete fixation at base
Evaluation criterion	Von Mises stress.

As can be seen in Equation (I), the equivalent density (ρ_{eq}) is calculated using the relative solid volume fraction (Vf), which is simply the complement of porosity (p).

The equation is strictly linear:

$$\rho_{eq} = \rho_s \cdot Vf \quad (I)$$

The value that defines the resin in Ansys will have a density ρ_{eq} for each TPMS structure.

To calculate the equivalent modulus of elasticity (E_{eq}) as a function of porosity (p), the material is modelled as a cellular solid, similar to a foam or lattice structure. In this context, the model used was the Gibson–Ashby, whose general expression is described by the following Equation (II):

$$E_{eq} = E_s \cdot C \cdot (Vf)^n \quad (II)$$

where C and n depend on whether the structure is dominated by bending or by tension/compression (such as certain lattice structures or oriented lattices). Our part will be dominated by compression. Since it is a resin with random porosity or a standard 3D printing infill, we use the most unfavourable values for the bending type. Above 90% porosity, the drop in stiffness is quadratic, not linear.

For random or open cell structures (sponge type), $C \approx 1$ and $n \approx 2$.

Unlike the density and modulus of elasticity, as the Poisson ratio of a porous cellular material does not scale with the porosity of the base material, depending almost exclusively on the pore geometry, we take a value of 0.33 obtained from the ANSYS additive manufacturing modulus.

2.2.5. Fabrication and post-processing

The five scaffold models with TPMS structures were produced by stereolithography (SLA) additive manufacturing on an AnyCubic Photon M3 Plus printer. The printing parameters are given in **Table 2**.

Table 2. Printing parameters

Parameter	Measurement
Layer thickness	0.05 mm
Lower layer count	6
Exposure time	3.2 s
Lower exposure time	42.05 s
Transition layer count	6
Type of transition	Linear
Reduction of transition time	5.55 s
Shutdown delay	0.5 s

The printed prototypes were subsequently cured for 8 minutes in an Anycubic Wash & Cure 3 Plus UV unit to complete post-polymerization and stabilize the fabricated structures. The same post-processing conditions were applied to all TPMS configurations.

2.2.6. Morphological analysis

A detailed morphological characterization of the fabricated scaffold structures was performed to evaluate the surface microstructure and quantify pore size and interconnectivity. Micrographs were captured from a Zeiss Stemi 508 stereomicroscope at 2× magnification with transmitted illumination, which provided detailed visualization of the internal topology. Pore measurements were obtained from Zeiss Labscope software.

3. Results

3.1. Reverse engineering

3.1.1. Segmentation of bone structure

Segmentation of the CT dataset enabled the maxillary bone to be isolated from the surrounding tissues. The reconstructed images revealed reduced bone volume in the left posterior maxilla, particularly in the region planned for the placement of two dental implants (**Figure 2**). The resulting anatomical model confirmed the extent and geometry of the defect used for the patient-specific scaffold design.

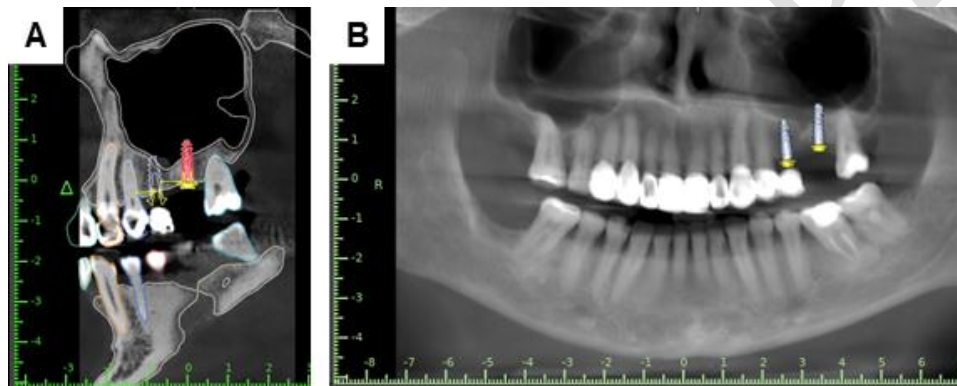


Figure 2. Segmentation of the bone structure in Blue Sky Plan® showing the position of potential implants. A) Sagittal section reveals reduced bone volume in the maxilla. B) Panoramic view of the tomography shows bone resorption in the posterior maxilla region.

3.1.2. 3D reconstruction

A digital 3D reconstruction was then used to generate a surface model of the patient's maxilla in STL format, including the provisional dental implants. This detailed model (**Figure 3**) served to visualize the exact morphology of the bone defect and confirmed the need for a bone regeneration scaffold.

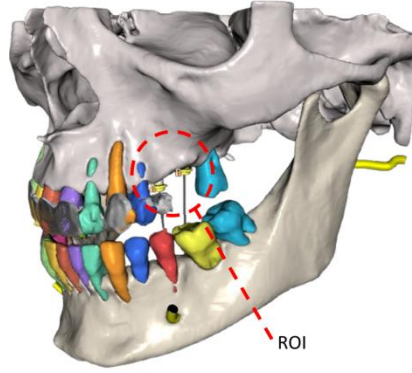


Figure 3. 3D reconstruction of bone structure, generated from the segmentation of the DICOM image, including dental implants, as can be seen in the region of interest (ROI).

3.1.3. Scaffold modelling

The specific volume to be regenerated was isolated on Meshmixer software and its geometry was reconstructed through a symmetry operation based on the patient's healthy contralateral side. This procedure yielded an anatomically accurate digital model of the bone missing from the maxilla. The resulting volume, shown in yellow in **Figure 4**, formed the geometric basis for the custom design of the scaffold intended to regenerate the defect.

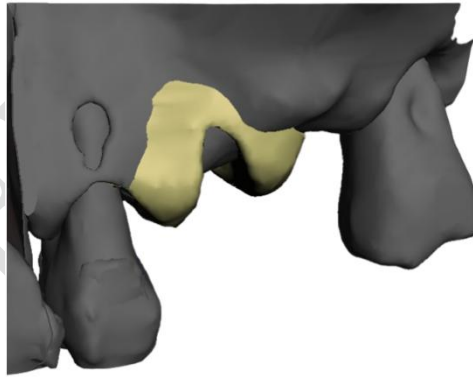


Figure 4. Modelling the bone volume to be regenerated by TPMS structures.

3.2. TPMS implementation

Five scaffold variants were designed, based on the anatomical model obtained using nTop™ software, each of which incorporated one of the TPMS structures with interconnected pores and different mechanical properties for bone regeneration. The cell size (1 mm) and wall thickness (0.16 mm) were kept constant for all the topologies to allow a direct comparison of their

performance. **Figure 5** shows the digital models of the scaffolds with the five different TPMS architectures.

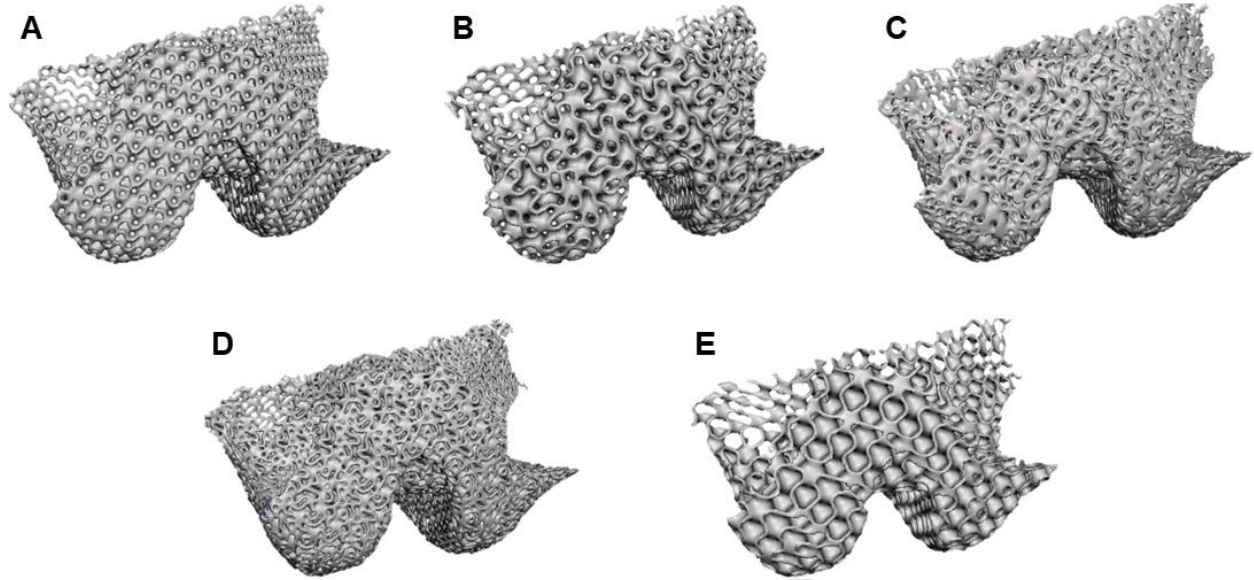


Figure 5. Personalized scaffold models with different TPMS structures: A) Diamond, B) Gyroid, C) Lidinoid, D) Split P, E) Schwarz P.

3.3. Mechanical evaluation

The model was prepared using a voxelized equivalent representation (**Figure 6**) to simulate these highly complex structures in ANSYSTM2025. This approach allowed us to assign homogenized material properties based on porosity and the solid volume fraction. The density associated with the different structures was calculated on the basis of these parameters.

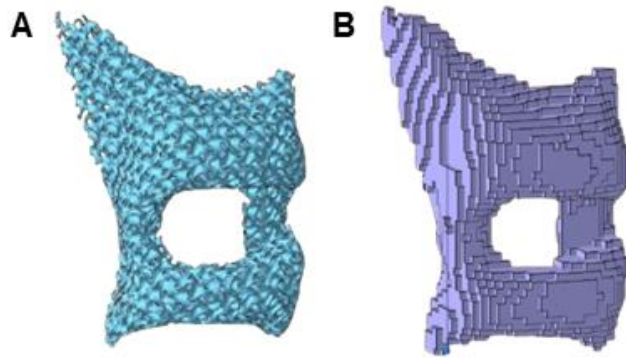


Figure 6. FE analysis models. A) Original lattice model of the TPMS structure. B) Voxelized equivalent model for the simulation of macroscopic properties.

The simulation of the voxelized model, carried out under the same loading conditions and boundary constraints, obtained the von Mises equivalent stress and total deformation values for each TPMS structure shown in **Table 3**. The only variation between the cases was in the apparent density properties according to the porosity of the different configurations.

Table 3. Results obtained for each TPMS topology under a compression load.

TPMS	Porosity (%)	Solid volume fraction (Vf)	Equivalent density (kg/m ³)	Equivalent modulus of elasticity (MPa)	Local Stress (MPa)	Deformation (mm)
Gyroid	93.13	0.069	81.2	7.56	3.83	0.00095938
Schwarz P	95.16	0.048	56.2	3.62	6.97	0.00020034
Diamond	91.51	0.085	98.4	11.1	2.51	0.00065335
Lidinoid	87.03	0.13	150.4	25.93	4.52	0.00027968
Split P	87.89	0.121	140.5	22.61	5.40	0.00032045

The FEA (**Figure 7**) showed that the Lidinoid and Schwarz P structures exhibited the least structural deformation, while the Gyroid structure showed the greatest structural deformation under the same load. Diamond and Split P exhibited intermediate behaviour.

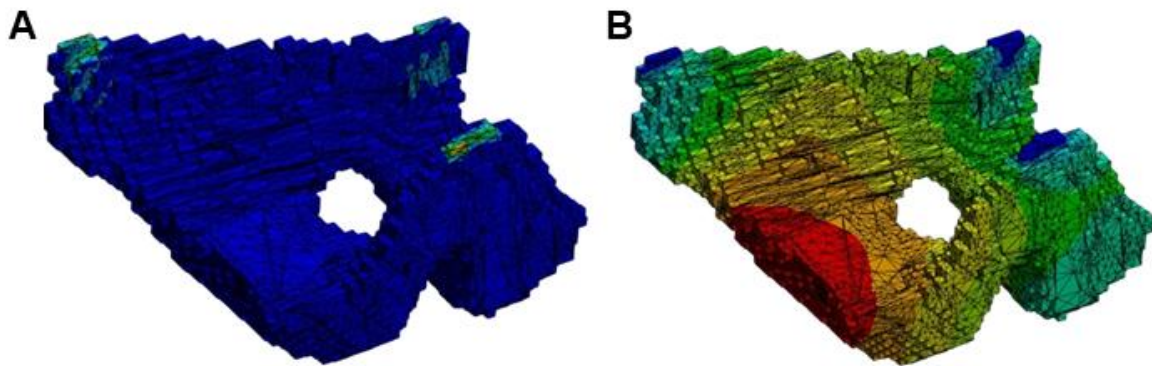


Figure 7. Results obtained in FEM of A) stress and B) strain of a voxelized equivalent Gyroid model with the application of the equivalent material.

Under identical loading and boundary conditions, the local von Mises stress was extracted separately for each TPMS topology at the location of maximum total deformation. These topology-specific values differ from the global nominal von Mises stress obtained from the homogenized voxelized representation.

3.4. Fabrication

The different TPMS structures were used to make five scaffold models, which fitted into the geometry of the patient's jaw, completing the bone structure required for the placement of dental implants, as can be seen in **Figure 8**.



Figure 8. Scaffold fabricated by SLA with Dental Model Water-Washable resin positioned on a patient-specific PLA bone replica.

3.5. Morphological analysis

Using a transmitted illumination setup, micrographs were captured at 2× magnification to evaluate the surface microstructure and quantify pore size. **Figure 9** gives the results for each structure.

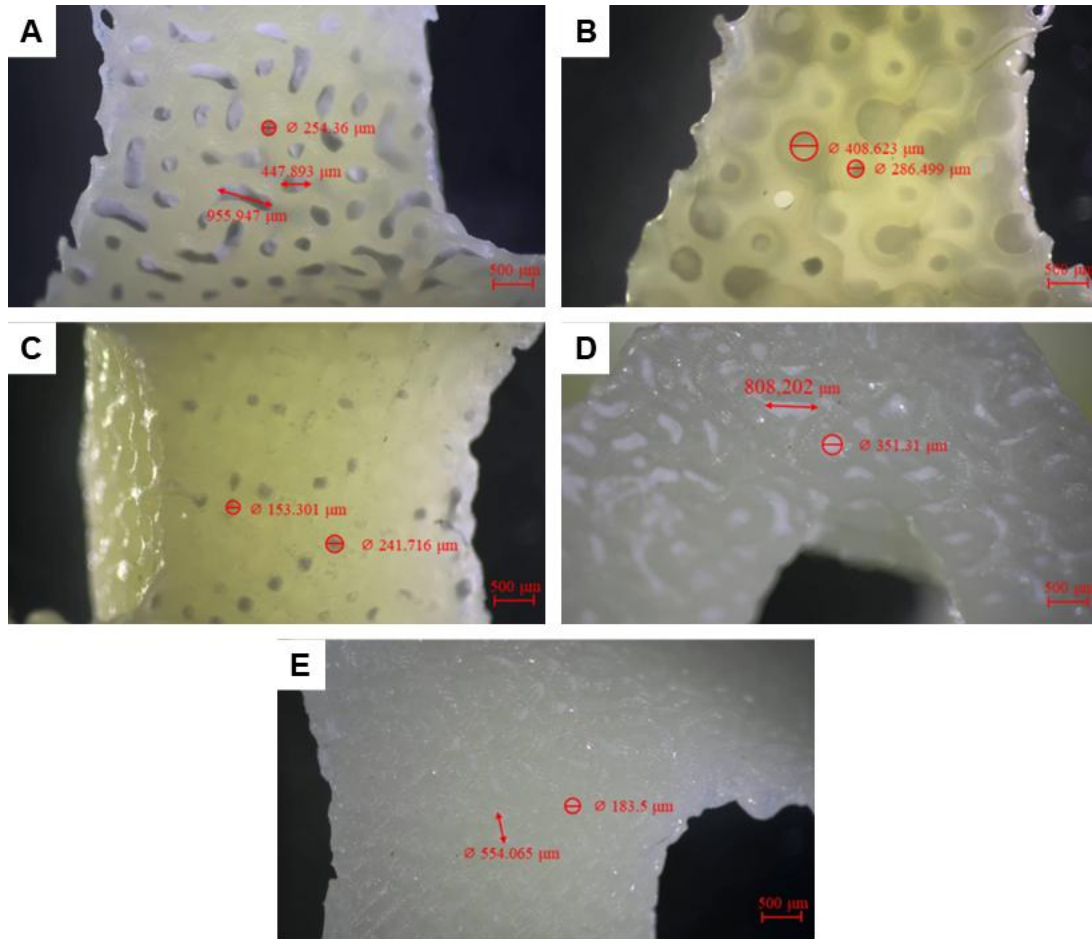


Figure 9. Micrographs of the designed tissue scaffolds. Structures: A) Gyroid, B) Schwarz P, C) Diamond, D) Lidinoid y E) Split P. Magnification 2 $\times$.

4. Discussion

The selection of the appropriate TPMS topology is a determining factor in the structural integrity and biological response of tissue-engineered implants. As revealed by the finite element simulation (FEA), the different TPMS architectures exhibited different mechanical behaviours under compression loading. Although the Lidinoid and Schwarz P structures showed the least deformation (indicating greater stiffness) excessive stiffness can be counterproductive in bone regeneration, in which the implant assumes a disproportionate share of the load, depriving the surrounding bone of the mechanical stimulus necessary for remodelling and growth. In contrast, the Gyroid structure exhibited the greatest deformation, which could mean a reduced capacity to withstand the initial functional loads. The Diamond and Split P structures, which showed a medium degree of mechanical behaviour in terms of deformation, offer a more promising balance

for biomimetic adaptation to native bone, thus promoting more successful and durable osseointegration.

The microarchitecture of the scaffolds is equally critical for biological functionality. Microscopic measurements confirmed the ability of additive manufacturing (SLA) to accurately replicate the complex pore curvatures and interconnectivity in TPMS structures. The pore sizes obtained (between 100 μm and 500 μm) are within the optimal range for promoting cell infiltration, nutrient diffusion and the vascularization essential for the formation of new bone tissue¹³. However, the pore uniformity varied significantly among the topologies. The Diamond structure exhibited the most homogeneous and well-ordered pore distribution, a crucial factor for guiding predictable and uniform cell growth, thus facilitating organized tissue regeneration. Although other structures such as Schwarz P also showed highly consistent pore sizes, Diamond's lower SD suggests a more controlled internal architecture, which is beneficial for the biological response. Despite its larger pores, the Lidinoid structure showed a less pronounced connectivity, which could be a hindrance to mass transport and cell growth.

Printing fidelity was influenced by the TPMS topology and its interaction with the selected SLA processing conditions. The descriptive pore-size analysis showed that 100% of the measured pores in the Diamond and Schwarz P scaffolds were within the target range of 100–500 μm , compared with 90.91% for Lidinoid, 86.67% for Gyroid, and 83.33% for Split P. Diamond also exhibited the lowest pore-size standard deviation, followed by Schwarz P, indicating a more uniform preservation of the visible open-pore morphology. In contrast, the larger dispersion observed for Gyroid and Split P suggests a less consistent reproduction of local pore openings under the selected fabrication and post-processing conditions.

The topology-dependent differences observed in the present study are consistent with previous investigations of photopolymer TPMS structures. Gabrieli et al.³¹ reported that stereolithography using a commercial dental resin could reproduce Gyroid and Schwarz P scaffolds with relatively small dimensional deviations, although scaffold morphology and compressive properties remained sensitive to wall thickness and relative density. Likewise, Peloquin et al.³² showed that the difference between designed and fabricated porosity depends on both lattice geometry and photopolymer properties, particularly resin viscosity. These findings support the interpretation that the pore-size variability observed in the present patient-specific scaffolds cannot be attributed

exclusively to the nominal TPMS geometry, but also to the interaction between topology, feature scale, material behaviour, and SLA processing conditions.

The observed differences may be related to topology-dependent surface curvature, orientation of local features relative to the build direction, resin drainage, and the sensitivity of thin walls to exposure and post-curing conditions. In the present study, the 1 mm unit-cell size and nominal wall thickness of 0.16 mm were close to the resolution limits imposed by the printing process. Consequently, small variations in polymerization or residual resin accumulation could produce comparatively large changes in pore opening and local wall continuity. Architectures with more regular and accessible openings, such as Schwarz P, may facilitate resin removal, whereas geometrically more complex internal pathways may be more sensitive to resin retention and partial pore closure. These interpretations should be considered explanatory hypotheses rather than directly measured manufacturing mechanisms.

The general SLA manufacturability of the same five TPMS families using the same type of resin and printing system was previously demonstrated by the authors³³. However, the standardized specimens used in that study had a unit-cell size of 7 mm and wall thicknesses ranging from 0.34 to 1.00 mm, whereas the present patient-specific scaffolds incorporated smaller cells and thinner walls. Therefore, the previous study supports the suitability of the general manufacturing approach but does not constitute a direct dimensional validation of the current anatomical scaffolds. Furthermore, the present stereomicroscopic assessment was limited to visible surface pore openings and did not quantify CAD-to-print surface deviation, actual wall thickness, volumetric porosity, internal connectivity, or dimensional shrinkage. Future studies should therefore include micro-computed tomography or three-dimensional scanning followed by registration with the original digital model to quantify these parameters.

Implant fixation, whether using calcium phosphate surgical cement or micro-screws, is essential to ensure the initial stability and promote successful osseointegration. This comprehensive strategy not only optimizes the bone regeneration process but also reduces the need for additional surgical interventions, resulting in lower risks, shorter recovery times and a lower cost for patients, thus improving their quality of life.

The Phrozen Dental Model Water-Washable Resin used in this study was selected exclusively to demonstrate the geometric design, SLA manufacturability, and morphological assessment stages of the proposed workflow. This rigid photopolymer is neither bioresorbable nor osteoconductive

and should not be considered suitable for permanent clinical implantation. Therefore, the fabricated structures represent proof-of-concept scaffold prototypes rather than clinically implantable bone-regeneration scaffolds. Translation of the proposed workflow toward clinical applications will require materials with appropriate biological, mechanical, and degradation properties, such as calcium-phosphate-based bioceramics, bioresorbable polymer–ceramic composites, or other osteoconductive composite systems^{7-10,25}. The adoption of these materials would also require re-optimization of the printing parameters, post-processing conditions, wall thickness, pore architecture, and mechanical response.

The numerical assessment was limited to a single static axial compression load uniformly applied to the superior surface of the scaffold. Although this condition provided a standardized reference for comparing the five TPMS configurations, it does not reproduce the full mechanical environment of a maxillary regeneration scaffold. Under clinical conditions, loads may be oblique, spatially non-uniform, cyclic, and may include shear components associated with mastication, implant position, fixation conditions, and patient-specific occlusal patterns. Therefore, the reported stress and deformation values should be interpreted as comparative predictions under an idealized loading condition rather than as a complete representation of in vivo mechanical performance. Future work should evaluate oblique and combined compression–shear loading, cyclic fatigue, and more realistic bone–scaffold contact and fixation conditions.

The exact patient-specific scaffold geometries were not subjected to experimental compression testing in the present study. Experimental characterization of the same five TPMS families, fabricated using the same type of resin and comparable SLA processing conditions, has previously demonstrated topology-dependent tensile and compressive behaviour^{31,33}. Nevertheless, those earlier tests were conducted on standardized specimens with different unit-cell dimensions, wall thicknesses, and boundary conditions. Consequently, they provide an experimental foundation for the selection and comparative interpretation of the TPMS architectures, but they should not be considered a direct validation of the numerical predictions obtained for the patient-specific maxillary scaffolds. Direct mechanical testing of anatomically shaped specimens is therefore required in future work.

The present study evaluated the patient-specific design workflow, comparative numerical response, SLA manufacturability, and visible surface pore morphology, but it did not assess biological performance. No tests of cytotoxicity, cell attachment, proliferation, osteogenic differentiation,

mineralization, osteoconduction, biodegradation, vascularization, or in vivo osseointegration were performed. Although a substantial proportion of the measured pores fell within a size range commonly associated with bone-tissue ingrowth, pore size alone does not demonstrate cellular compatibility or biological functionality²⁵⁻²⁸. The absence of biological validation therefore represents a significant limitation of this proof-of-concept study. Future work should include in vitro cytocompatibility, cell adhesion, proliferation, and osteogenic differentiation studies, followed by degradation testing and in vivo evaluation of vascularization, bone formation, osseointegration, and long-term host response using clinically suitable bioresorbable and osteoconductive materials (Table 4).

Table 4. Statistical analysis of pore sizes.

Structure	Mean	Median	Standard Deviation (SD)	% pores between 100 μm and 500 μm
Gyroid	418.55	378.80	166.61	86.67
Schwarz P	264.30	238.73	57.92	100.00
Diamond	126.05	120.77	25.84	100.00
Lidinoid	282.96	257.82	108.22	90.91
Split P	420.21	383.90	154.55	83.33

5. Conclusions

This study presents a reverse-engineering methodology for the patient-specific design of TPMS-based scaffold prototypes for maxillary bone regeneration. Tomographic images from an actual clinical case were used to segment the affected maxillary region, reconstruct the bone defect, and generate an anatomically adapted three-dimensional scaffold model.

Five TPMS topologies; Gyroid, Schwarz P, Diamond, Lidinoid, and Split P, were implemented using the same unit-cell size and wall thickness and were subsequently assessed through a comparative finite element analysis, stereolithographic fabrication, and morphological

characterization. The SLA process enabled the fabrication of all five patient-specific scaffold configurations, while the morphological analysis showed that a high proportion of the measured pores fell within the 100 - 500 μm range. Diamond and Schwarz P exhibited the greatest pore-size uniformity among the evaluated structures.

Within the limits of the comparative numerical assessment, the Diamond topology showed a favorable combination of predicted local stress response and measured pore-size uniformity. This result supports its selection as a candidate for subsequent experimental and biological validation but does not constitute direct evidence of clinical load-bearing performance.

The resin-printed structures should be interpreted strictly as proof-of-concept prototypes and were not intended for clinical implantation. Translation of the proposed workflow will require the use and validation of osteoconductive and bioresorbable bioceramic, polymeric, or composite materials.

Overall, the proposed workflow demonstrates the feasibility of integrating medical image reconstruction, implicit TPMS modelling, finite element assessment, and stereolithographic fabrication for the development of patient-specific scaffold prototypes. Direct mechanical testing of the anatomically shaped scaffolds, together with in vitro and in vivo studies, is still required to validate their compressive behaviour, biological response, osseointegration, and potential clinical applicability.

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Conflict of interest

The authors declare they have no competing interests.

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Ethics approval and consent to participate

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Not applicable.

Availability of data

Data are available from the corresponding author upon reasonable request.

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