

RESEARCH ARTICLE

Design and manufacturing of a 3D-printed field-driven metamaterial pelvic bone plate

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Abstract

Conventional pelvic bone plates have inherent limitations, including poor anatomical matching, unsatisfactory mechanical performance, excessive weight, and inadequate biocompatibility. This study aims to develop a personalized pelvic bone plate by integrating field-driven metamaterial design with three-dimensional (3D) printing. The methodology involves computed tomography (CT) and Mimics software for reconstruction of the pelvic model, finite element simulation for topology optimization, and 3D printing for direct fabrication of the designed plate. The results demonstrate that the bone plate reconstructed from the pelvic fracture surface exhibited high anatomical fit. After topology optimization, the deformation of the bone plate increased by approximately 15%, accompanied by uniform stress distribution without obvious stress concentration. For the field-driven porous bone plate, the relative density of the porous structure increased with increasing local stress. The partially porous bone plate achieved a 30.77 % weight reduction while combining the favorable mechanical properties of solid plates with the biological potential of porous architectures. The 3D-printed personalized bone plate exhibited excellent forming quality, surface finish, and assembly fit. This study demonstrates the feasibility of the proposed structural design and 3D printing process, providing theoretical support and technical references for the subsequent development of high-performance personalized pelvic bone plates.

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Keywords: 3D printing; Pelvic bone plate; Topological optimization; Porous structure; Manufacturing quality

1. Introduction

A recent study published in *The Lancet Health Longevity* showed that there were about 455 million fractures worldwide in 2019 (an increase of 70.1% over 1990) and about 178 million new fractures (an increase of 33.4% over 1990).¹ Internal fixation with bone plates treats bone diseases, trauma, congenital malformations, and other diseases. The plate's performance is a critical factor in the success of osteosynthesis.^{2,3} Most scholars believe that the two primary factors affecting the performance of bone plates are

materials and structural design.^{4,5} The study of material factors is relatively mature. Common plate materials include Ti6Al4V and 316L alloys. The structural design of the bone plate is the focus of current research. Regarding structural design, the plate screws should be kept to a minimum to avoid stress concentration.

Moreover, the inner surface of the bone plate should demonstrate good fit, be lightweight, and exert small stress concentrations on human bone. Current plates are prone to migration and stress concentration after implantation because they are manufactured in bulk. With improvements in living standards, there are greater demands on the performance of bone plates. Biofixation plates (partially or fully porous structures) have gained appreciation due to their low stress concentration, less susceptibility to displacement after implantation, and excellent biocompatibility.⁶ However, their complex structure makes manufacturing using traditional processing methods challenging.

The advent of three-dimensional (3D) printing technology has enabled the manufacture of biological fixation plates, a high-performance, personalized, and complex type of bone plate. These technologies use specialized software to obtain cross-sectional data by slicing and layering the 3D model, and then import these data to rapidly shape the device and fabricate entity components by a layer-by-layer accumulation of materials. Utilizing the layer-by-layer accumulation method, 3D printing technology can almost fully manufacture geometric-shaped parts by processing single parts, small batches, and parts with complex geometric structures.⁷⁻⁹ Selective laser melting (SLM) is a 3D printing technology that uses a laser to melt metal powders.¹⁰

Wei *et al.*¹¹ explored the significance of 3D-printed personalized plate internal fixation in treating severe tibial plateau fractures. Wang and Wang¹² investigated the effect of different gaps between the plate and the bone surface on the acetabular stress distribution after treating posterior acetabular wall fractures with internal plate fixation using a 3D finite element method. By establishing a simplified finite element model of the tibial bone plate, Zhang *et al.*¹³ combined finite element and data-sampling methods to simulate the solid and fractional bone plate systems, respectively. The lightweight design of the bone plate was achieved by comparing its mechanical properties and improving its resistance to bone stress. Shi *et al.*¹⁴ studied the effect of digital design combined with 3D printing technology in the adjuvant surgical treatment of complex tibial plateau fractures. Pobloth *et al.*¹⁵ designed honeycomb 3D titanium alloy mesh scaffolds with varying stiffnesses and compared them for bone growth and healing after

changing the scaffold stiffness; the honeycomb titanium alloy mesh scaffold reduced the shielding effect of stress and promoted the healing and regeneration of large-animal bones. Vijayavenkataraman *et al.*¹⁶ proposed a new bone plate design method that combines two auxetic structures (a reentrant honeycomb structure and a missing rib structure) and compared it with traditional plate designs; plates with reentrant honeycomb structures performed well. Mie *et al.*¹⁷ created a 3D-printed, customized Ti-6Al-4V plate that fits the shape of the dog's radius and placed it at the site of the osteotomy. Liu *et al.*¹⁸ used 3D printing to prepare personalized porous Ti6Al4V plates, which were subsequently coated with Ta to achieve elastic moduli similar to those of cortical bone; no stress shielding was observed. Based on computed tomography (CT) images, Yao *et al.*¹⁹ designed bone plates that matched bones by constructing patient-specific bone models manufactured using 3D printing; recovery significantly improved after the plates were implanted in the patient's foot. Relevant progress has also been made in composite scaffolds.²⁰ Günther *et al.*²¹ optimized the triply periodic minimal surface (TPMS) lattice geometry. The results indicated that the maximum stiffness improvement reached 80% and the corresponding strength improvement reached 61%, while preserving the intrinsic morphological characteristics of TPMS. El-Sayed *et al.*²² conducted an experimental design using response surface and variance analysis to investigate the influence of design parameters on the performance of additive manufacturing lattice structures, such as ultimate compressive strength, specific compressive strength, elastic modulus, and porosity. Al-Tamimi²³ designed a new type of bone plate, and results showed that as volume increased, load transfer in the fracture area decreased, thereby reducing the risk of stress shielding. Strömberg²⁴ derived the functional gradient of the TPMS lattice structure based on topological optimization.

The design and manufacture of new bone plates have become a research hotspot. However, most research focused on analyzing lattice structures and modifying bone plate surfaces. Plate performance remains to be improved. Previous studies have reduced stress shielding by 20–40% through lightweight and porous designs, but they still lack a balanced consideration of mechanical performance, manufacturability, and biological function. Few studies have introduced field-driven adaptive design into personalized pelvic plates. Thus, further optimization is still needed. Therefore, parametric modeling was combined with the finite element method in this study to carry out lightweight design of biological fixation bone plates (partially or fully porous structures) with modulus-conforming characteristics. The modulus adaptation is achieved by designing a reasonable pore size, porosity, and

a graded porous distribution, thereby reducing the elastic modulus of the bone plate to be close to that of human cortical bone. The effect of modulus matching is further evaluated by comparing the elastic modulus, interfacial stress distribution, and overall stress shielding level with those of the traditional solid bone plate.

2. Materials and methods

2.1. Design constraints

The bone plate was designed according to the medical device industry standard YY 0017-2016 for metallic bone plates for osteosynthesis implants.²⁵ The design should also meet the following three constraints:^{26,27} (i) sharp angle and thin-wall constraints: the minimum size of the processed bone plate should be greater than the minimum spot diameter; the laser spot diameter of GYD150 SLM equipment (Sunshine Laser & Electronics Technology Co., Ltd., China) should be 50–70 μm , and the minimum structural size of the bone plate should be larger than 100 μm ; (ii) hole constraints: the minimum pore size of the designed bone plate should be greater than the laser spot diameter, and the effect of powder adhesion should also be considered during processing; and (iii) biocompatibility constraints: the pore size range of the porous structure in the designed bone plate should be 100–1,000 μm , the porosity should be 50–90%. A larger surface area-to-volume ratio is preferred.

2.2. Materials and manufacturing methods

To reduce costs and improve manufacturing efficiency, an industrially high-precision desktop 3D printer, the Z-603S (Shenzhen Sunshine Laser & Electronics Technology Co., Ltd.), was used to print the host bone (pelvis). The material was polylactic acid (PLA), the printing layer thickness was 0.15 mm, the packing density was 18%, the wall thickness was 0.6 mm, and the printing temperature was 190°C. To avoid warping deformation caused by an excessive temperature difference between molded parts and the base plate, the base plate was preheated to 50 °C.

The pelvic bone plate was made of Ti6Al4V alloy metal powder (Falcontech Rapid Manufacturing Technology Co., Ltd., China). Its compositions meet ASTM F136 and GB/T 13810-2007 requirements. The compositions are shown in Table 1. The powder was prepared by gas atomization and was spherical, with a loose packing density of 2.55 g/cm³. The particle size distribution was concentrated in a narrow region, with D90 of 57.9 μm and D50 of 35.6 μm .

The GYD150 SLM equipment (Sunshine Laser & Electronics Technology Co., Ltd.) was used for the molding. Argon was used as a protective gas. Oxygen content was controlled below 0.03%. The laser processing

power was set to 180 W, the scanning speed to 500 mm/s, the scanning spacing to 80 μm , and the processing layer thickness to 30 μm . A scanning strategy with X–Y interlayer interdigitation was employed.

Table 1. Comparison of powder material manufactured in SLM and the ASTM F136 standard

Element	Ti6Al4V powder	ASTM F136 Standard
Al	5.5–6.5%	5.5–6.5%
V	3.5–4.5%	3.5–4.5%
Fe	0.25%	<0.3%
C	0.08%	<0.08%
N	0.03%	<0.05%
H	0.012%	<0.012%
O	<0.08%	<0.13%
Ti	Balance	Balance

Abbreviation: SLM: Selective laser melting.

2.3. Analysis methods

The pelvic model used in this study was obtained from the built-in demonstration sample dataset of Mimics 22.0 software (Materialise, Belgium), not from clinical patient images. These sample models are designed for teaching and academic research. Accordingly, ethical approval and informed consent were not required. A single typical pelvic model was adopted to verify the feasibility of the proposed design and manufacturing method. First, the pelvic bone was reconstructed in reverse using Mimics 22.0. Then, the data were imported into Geomagic Design X 2020 (Geomagic, United States of America [USA]) for fracture reduction and solidification. Next, the positive design of the bone plate was carried out in Rhino 6.0 (Robert McNeel & Assoc., USA). Finally, the completed bone plate design was imported into Inspire 2022 (Altair, USA) for finite element analysis and topological optimization design. The porous structure filling was conducted in Rhino, while data processing and risk analysis for 3D-printed parts were conducted in Magics 22.0 (Materialise).

Blasting was performed first, followed by polishing with sandpaper. Finally, parts were polished using a polishing cloth. A high-definition video graphics array (VGA) electron microscope (BC200VGA-A1, Shanghai BOCHENG Co., Ltd., China) was adopted to characterize the surface morphology of the 3D-printed parts after surface treatment. SLM molding process parameters were optimized to obtain the optimal molding parts.

3. Results and discussion

3.1. Reconstruction and simulation of pelvic repair

3.1.1. Reverse reconstruction of the pelvic model

The fracture site is usually located on a complex geometry or curved surface. Therefore, to improve the fit between the designed bone plate and the affected anatomical region, reduce the risk of plate fracture, and increase the success rate of osteosynthesis surgery, imaging data were acquired using CT and *magnetic resonance tomography* (Figure 1a). The pelvic data from the CT scans were then imported into Mimics. The pelvic bone model was obtained by thresholding, region growing, morphological manipulation, 3D reconstruction, and surface processing, as shown in Figure 1b.

3.1.2. Pelvic plate design

Pelvic models are complex, and blocking may occur if the plate design is performed directly on the model. The amount of data is also significant. Therefore, models other than the disease site may be removed before design. We first imported the reconstructed pelvic bone model into 3-Matic 12.0 (Materialise) to remove all parts except the disease site. The patient part was divided, and the segmented patient part was stitched. The plate placement was marked, as shown in Figure 2a. After smoothing the marked edges and separating the curved surface of the plate, an offset thickening of 2.5 mm was applied to meet the design requirements, yielding the original plate model, which was saved in STL format, as shown in Figure 2b. Finally, the saved STL-format plate model was imported into Design X to fix triangular errors using Mesh Physicians. Substantialization was performed. Ten circular holes with different depths and a diameter of 3.5 mm were opened above it to facilitate the temporary unthreading in the simulation hole, as shown in Figure 2c and 2d. For

such complex pelvic fractures, at least three plates are required when using traditional bulk plates. In contrast, if a personalized plate is used, only one is needed, and the plate surface and the osteosynthetic surface complete the fit, significantly improving the procedure's success rate.

3.2. Finite element analysis of pelvic bone plate

3.2.1. Simulation parameters for bone plates

The 3D solid plate model was saved in IGES format and imported into Inspire software for force analysis. The material was set to Ti6Al4V (elastic modulus 116.52×10^3 MPa, Poisson's ratio 0.31, and density 4.429 g/cm^3). The overall length of the pelvic bone plate model was 59.37 mm, the span was 48.29 mm, and the average thickness was 2.52 mm, as shown in Figure 3a. Thickness deviation is inevitable due to the complex curved surface of the plate, making accurate measurement difficult. The pelvic bone was subjected to complex loading conditions in humans. There were multiple influencing factors, including screw pre-tightening, friction, and force between muscles, ligaments, and the plate. Stretching, bending, and torsion were also considered. The femoral head gave the ischium and ilium a specific reaction force when compressed, which was dominated by tensile force when transmitted to the plate. Therefore, bending and torsion did not need to be applied separately. The finite element analysis was used for preliminary design comparison and topology optimization, not for a reliable assessment of clinical fixation stability. For an adult of 70 kg, a load of 525 N was applied to each of the four holes at the front end, with a total load F of 2,100 N (about three times the body weight). Six holes at the other end of the plate were set for complete fixation, as shown in Figure 3b. The mesh type was set to tetrahedral, with a unit size of 1 mm. The number of mesh elements was 23,650. This mesh size was sufficiently refined to ensure mesh convergence and stable numerical

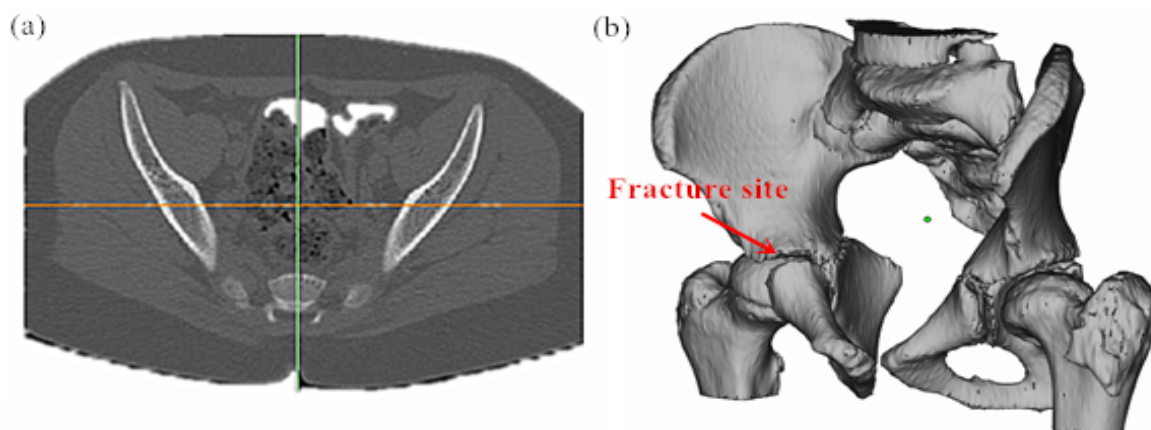


Figure 1. Three-dimensional reconstruction of the pelvis. (a) Computed tomography data of the pelvis. (b) Three-dimensional models of the pelvis.

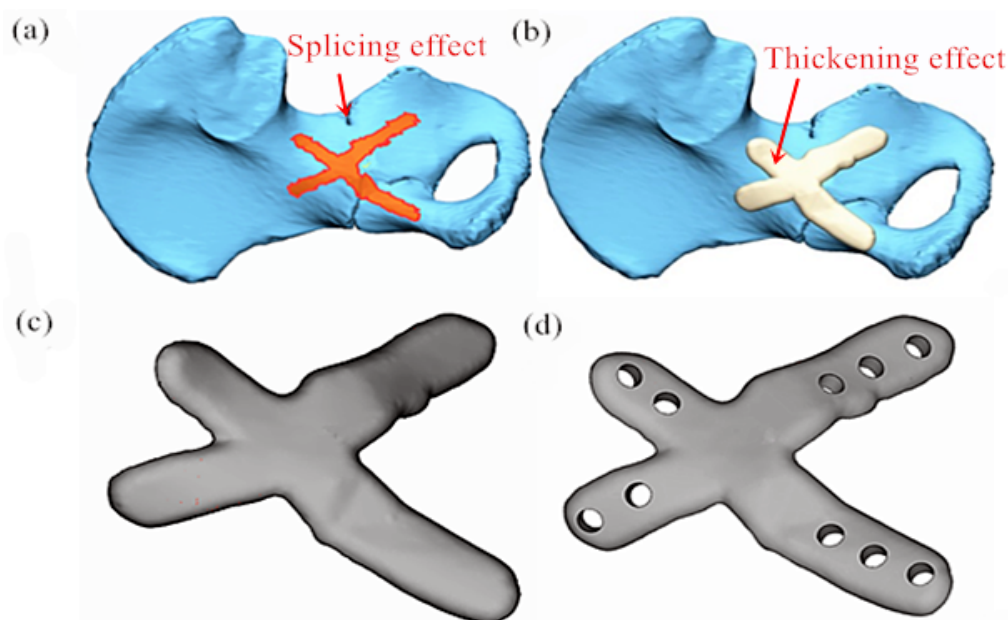


Figure 2. Plate repair of pelvic fracture. (a) Marking of the placement position of the plate. (b) Design of the bone plate. (c) Solidification of the bone plate. (d) Opening of the screw hole of the bone plate.

solutions. This study focuses on conceptual structural design, topology optimization, and relative mechanical performance comparison, rather than high-precision absolute numerical calculation. For this type of preliminary engineering simulation, the adopted refined mesh meets the conventional analysis requirements. The mesh division result is shown in Figure 3c. The OptiStruct solver was employed for numerical computation. The static analysis was performed using the Gaussian elimination method, while the topology optimization adopted the DUAL2 algorithm. A dual convergence criterion was implemented, allowing termination if either criterion was satisfied.

3.2.2. Simulation result analysis of bone plates

Finite element simulation results for the pelvic plate are shown in Figure 4. The displacement of the pelvic plate decreased from left to right, showing a gradient change, as shown in Figure 4a. Maximum displacement occurred at 0.19 mm in the upper left head of the plate. A small amount of deformation met the use requirements. The stress primarily occurred in the middle of the plate, as seen in Figure 4b. The changing trend was observed from the middle to both ends, and the stress concentration was small, improving the traditional plates' stress concentration phenomenon.

The maximum stress was located at an angle of 451 MPa along the length direction of the bone plate, which was lower than the yield strength of the TC4 material at 825–895 MPa. The minimum safety factor was 1.9, which

meets the safety requirements and has certain optimization space. The mass of the pelvic plate (Ti6Al4V material) was 13 g. Further plate optimization may be performed to reduce the weight and improve the biocompatibility of pelvic plates.

3.3. Topology optimization for pelvic plates

3.3.1. Design parameters for topology optimization of bone plates

Simulation parameters, including material properties, loading conditions, and constraints identical to those of the solid plate, were defined in Inspire. To ensure that the screw hole edges of the plate were fully completed after topological optimization, the screw hole edges could be divided. For this purpose, we used a segmentation command in Inspire to segment the screw holes with a surrounding thickness of 0.3 mm. Subsequently, the plate was designated as the design space, with the exception of the screw holes (the dark red regions in Figure 5). Due to the irregular shape of the pelvic plate, the topologically optimized shape constraints were left out of control. Finally, we set the plate topology optimization quality target to 30%. The plate thickness constraint was set to 1 mm, considering the laser-printed part constraints and computational efficiency. Plate constraints and loads applied to the pelvis are shown in Figure 5.

3.3.2. Topology optimization of bone plates

The results of the topological optimization of the pelvic plate

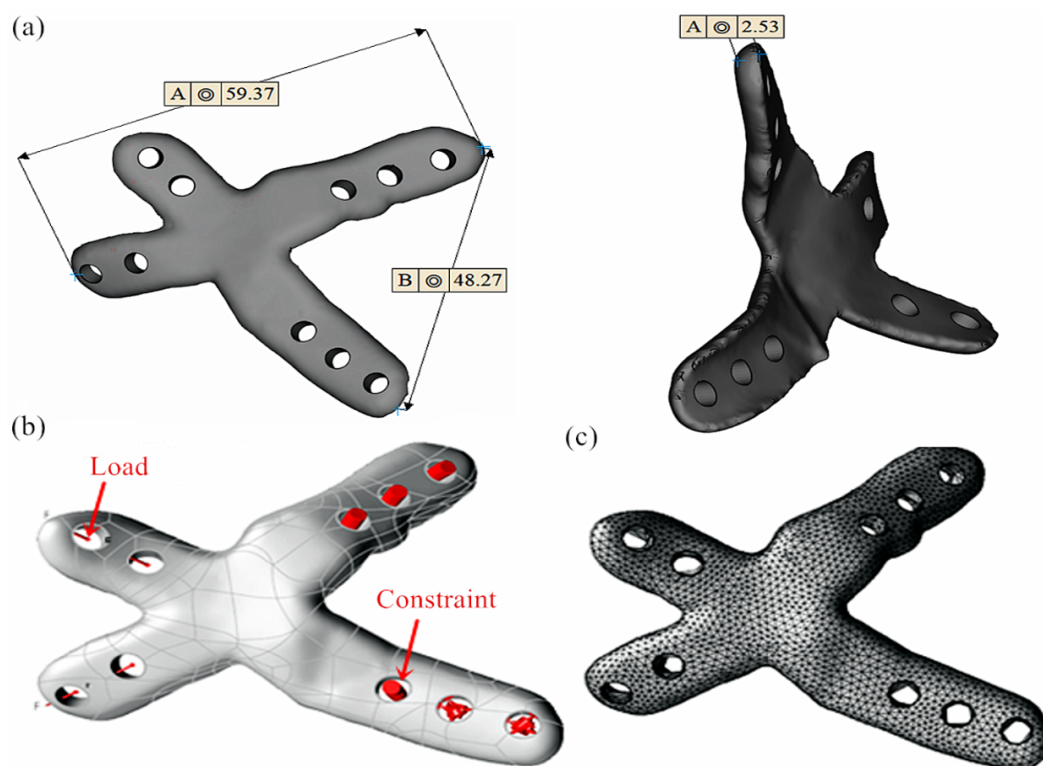


Figure 3. Simulation parameters of the bone plate. (a) Plate size information. (b) Schematic diagram of load and constraint application. (c) Meshing results.

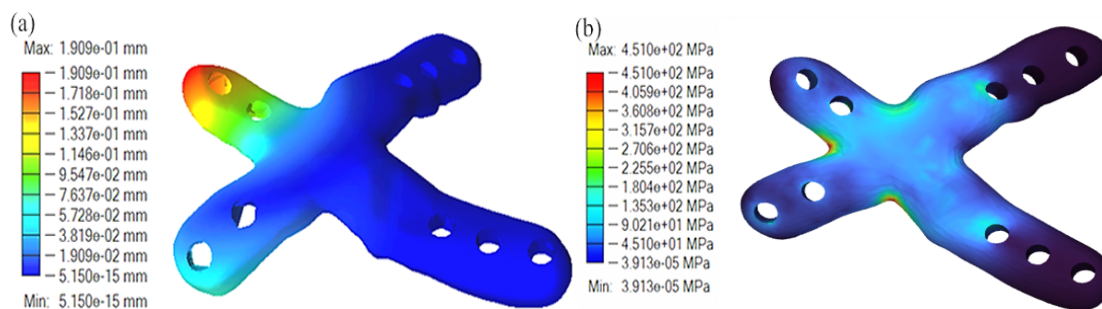


Figure 4. Finite element simulation results of the pelvic plate. (a) Displacement contour plot. (b) Stress contour plot.

are shown in [Figure 6](#). The surface of the plate completed by topological optimization was smooth, and the stress hole and the fixed hole were connected through the plate, as seen in [Figure 6a](#). Fixation holes with less stress on the tail were not connected and could be removed. The displacement of the pelvic plate gradually decreased from left to right, and the trend of displacement change was similar to that before topology optimization, as seen in [Figure 6b](#). The maximum displacement occurred at the superior left head of the plate at 0.22 mm. The deformation amount increased by 15% compared to before optimization, but it was still relatively small and met the requirements for use. The stress distribution of the plate completed by topology

optimization was similar to that before optimization, and the stress concentration was slightly increased, as seen in [Figure 6c](#). The maximum stress was 6.42×10^2 MPa. No stress concentration points were evident, ensuring the plate did not break abruptly. Poly-non-uniform rational B-spline (PolyNURBS) fitting was performed on the pelvic plate model and completed with force analysis using Inspire, as shown in [Figure 6d](#). The curved surface of the bone plate had the same size and uniform distribution, as shown in [Figure 6d](#). No prominent deformation surfaces were present. The fixation holes combined well with the plate and met the use requirements. The mass of the pelvic plate was 4.09 g, representing a 68.54% reduction

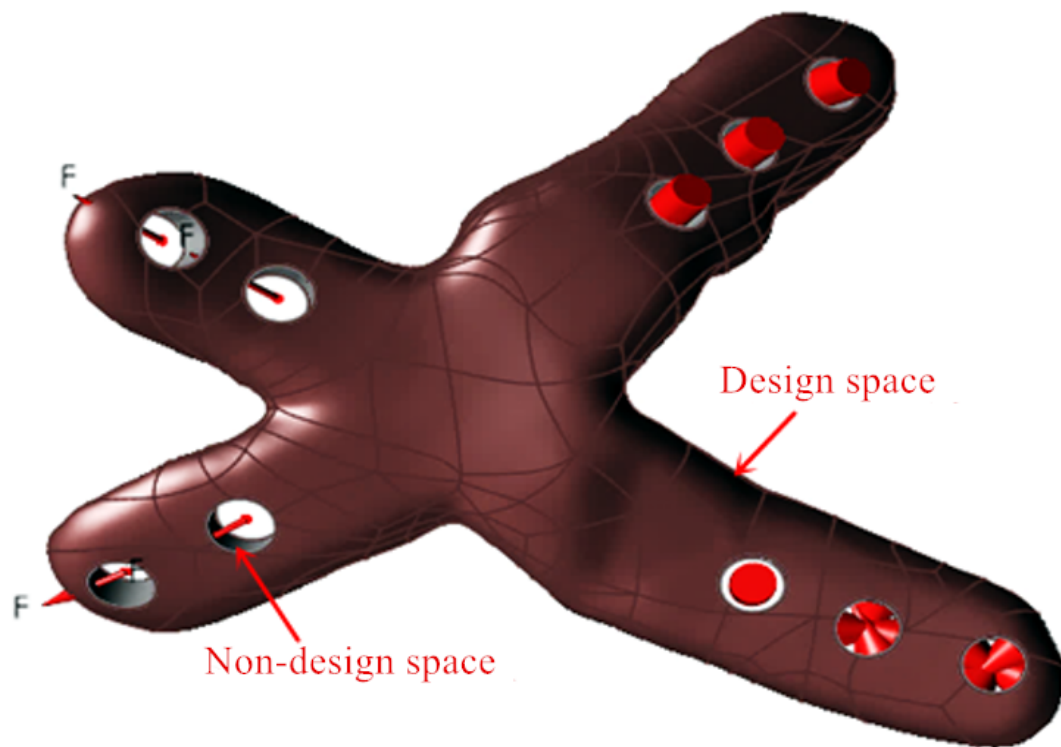


Figure 5. Schematic diagram of the restraint and load application of the pelvic plate

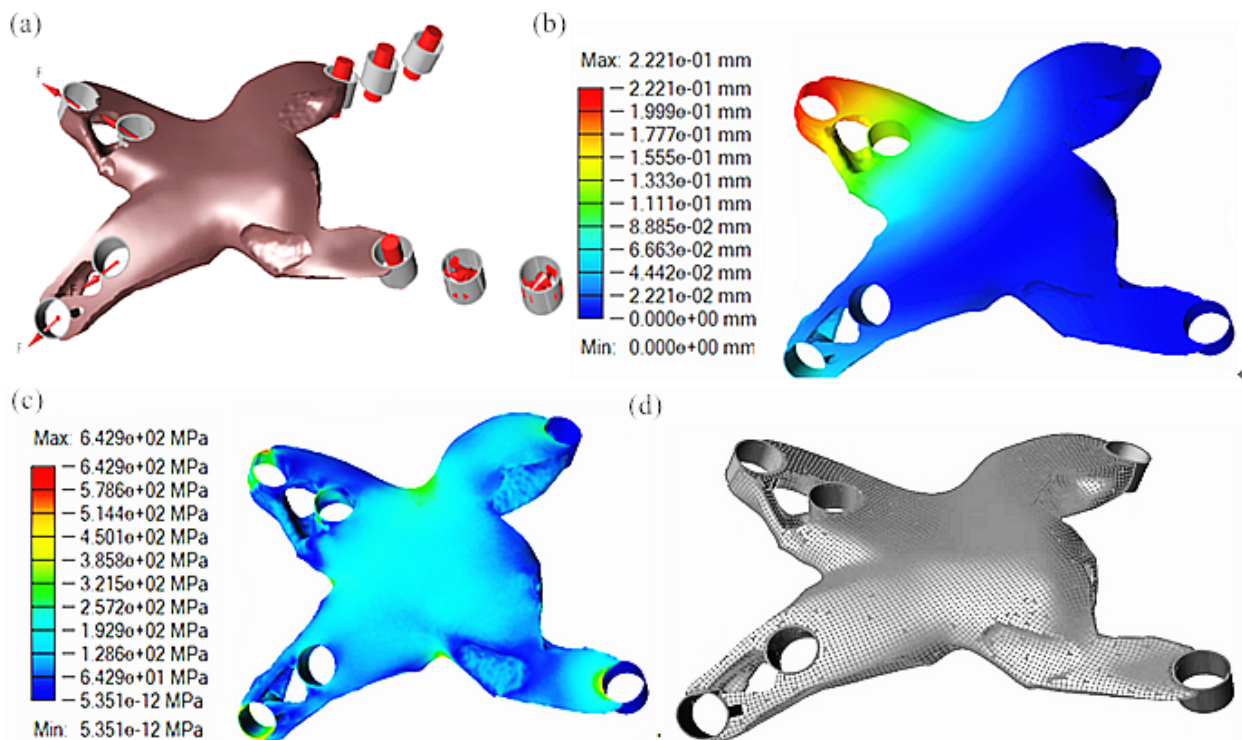


Figure 6. Results of topology optimization and force analysis of the pelvic bone plate. (a) Topology optimization effect. (b) Displacement contour plot. (c) Stress contour plot. (d) Fitting effect.

compared with the model before topological optimization, achieving the weight reduction goal. In order to improve the biomechanical behavior of the bone plate, further design can be carried out.

3.4. Biological fixation design of pelvic bone plate

3.4.1. Partial porosity of plate

When designing the biological fixation pelvic bone plate, the application range of different porous structures must be considered, including whether the bone plate fits closely with the physical part after filling with a porous structure, whether it meets the processing requirements, and whether it is conducive to the improvement of the mechanical properties and biocompatibility of the pelvic bone plate. For this reason, a method of simulating the filling of the porous structure was used to select a porous structure suitable for pelvic plates.²⁸ First, porous structure units R (rhombic dodecahedron with a relative density of 20%), G (gyroid), and B (body diagonals with nodes rounded) with excellent mechanical properties were selected. Compared with conventional TPMS porous structures, these three lattice units feature simpler strut connections, better structural stability, and higher compatibility with SLM manufacturing, making them more appropriate for the complex, curved shape of pelvic bone plates. Then, according to the rules of porous structure generation, the topologically optimized pelvic plates were simulated using parametric programming in Rhino 6. The results are shown in Figure 7; the topologically optimized plate entities filled with different elements fit closely with the porous parts. The porous structure parameters (porosity, mean pore size, and surface area-to-volume ratio) met the implantation requirements; however, there were differences in processability among porous structures. As seen in Figure 7, when the R unit was placed according to the optimized placement mode, the included angle with the machining platform was 36°, which increased the risk of machining failure (to ensure the quality of parts, it was required to be no less than 45°); some struts of the G unit porous structure protrude longer, extending about 0.32 mm in the parallel position with the platform. We found that the G unit collapsed during actual processing; the porous structure generated by the B unit exhibited bulging. However, the bulges were minor and did not affect processing. In addition, the porous structure struts generated by the B unit were well bonded to one another, ensuring processing quality. Therefore, when the mechanical properties and biocompatibility of the three structural units met the requirements, the more suitable B-unit porous structure for processing was selected to generate the biological fixation pelvic bone plate. The final mass of the pelvic bone plate uniformly filled with the B-unit porous structure

was 9 g, resulting in a 30.77% weight reduction compared with the original structure before optimization. All porous structures in this study were designed to meet the pore size and porosity requirements in Section 2.1. Direct measurement of specific parameters was not possible due to the complex and irregular pore distribution. Screws with a diameter of 3.5 mm were designed according to the ASTM-F543-07 Standard Specification and Test Method Standard for Metallic Medical Bone Screws. Then, the designed screws were imported into the software, which performed Boolean operations on the plate to open the screw holes after placement adjustments. The B unit was mainly adopted for uniform porous filling. For the subsequent stress-field-driven gradient porous design, we selected the diamond structure for its better topological flexibility and adaptive regulation capability.

3.4.2. Full porosity of pelvic plate

(a) Fully porous plate based on topology optimization

Solid plates were imported into the Inspire software for porosity analysis. Simulated parameters, including material, load, and constraints, were defined for the topologically optimized plates. Optimization was performed by setting the target length of the porous structure to 3 mm, the minimum diameter to 0.2, and the maximum diameter to 0.7 to maximize stiffness. The results of the porous structure simulation of the plates are shown in Figure 8. The displacement variation trend of the plate was similar to that before optimization, as shown in Figure 8a. The maximum displacement was 0.74 mm, 3.8 times larger than before optimization, and the deformation was marginally significant. The stress distribution of the porous plates was similar to that before optimization and concentrated in the middle, as shown in Figure 8b. The maximum stress was 3.56×10^3 MPa, 7.8 times higher than before optimization. The weight of the pelvic plate was approximately 5 g, which is 61.54% lighter than before topological optimization. The minimum safety factor was 0.05 (i.e., small). This indicates that the fully porous design failed mechanical verification under the tested loading condition and was not mechanically feasible as a primary load-bearing implant for the pelvis. According to previous studies, fully porous structures may enhance biocompatibility and tissue ingrowth, which are well-recognized advantages of porous orthopedic implants.²⁸ The lower elastic modulus matches natural bone better and helps reduce stress shielding. However, the stress concentration and deformation increase as well. Therefore, fully porous bone plates are not suitable as a standalone fixation for high-stress pelvic sites, but may still be used as local components in low-stress regions, such as the ulna, to prevent stress concentration.

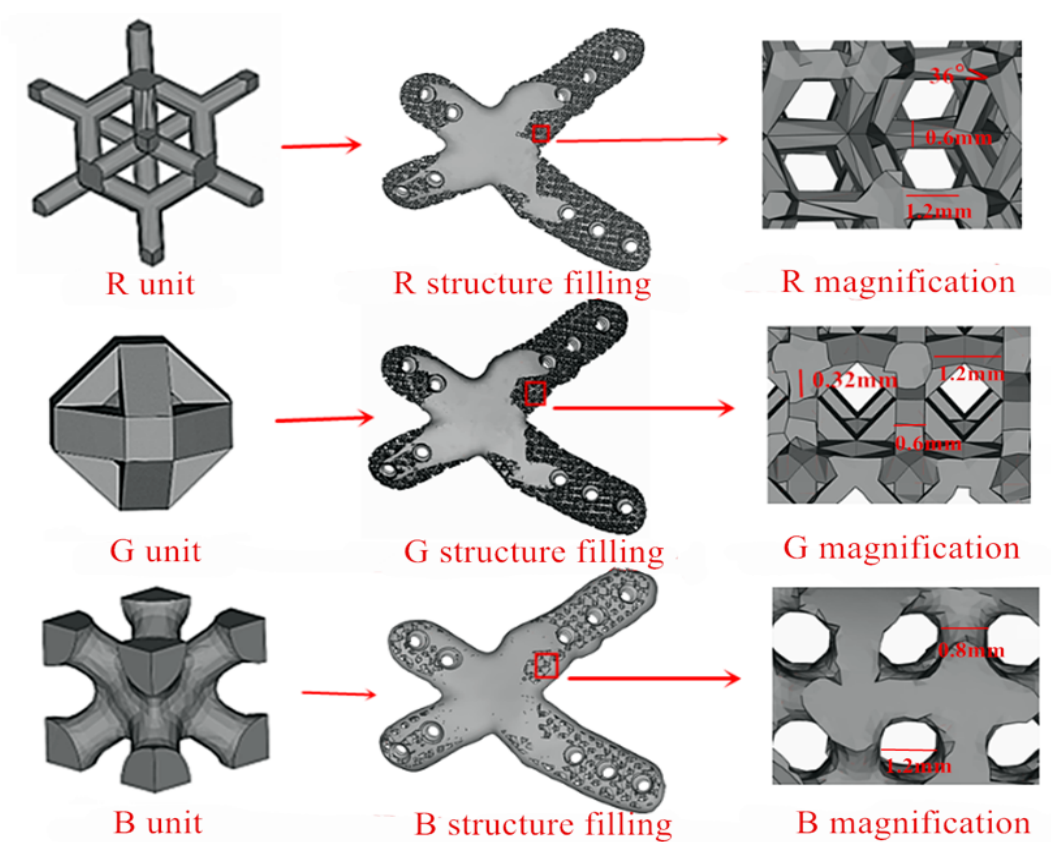


Figure 7. Effects of topology-optimized bone plate filled with different porous structural units

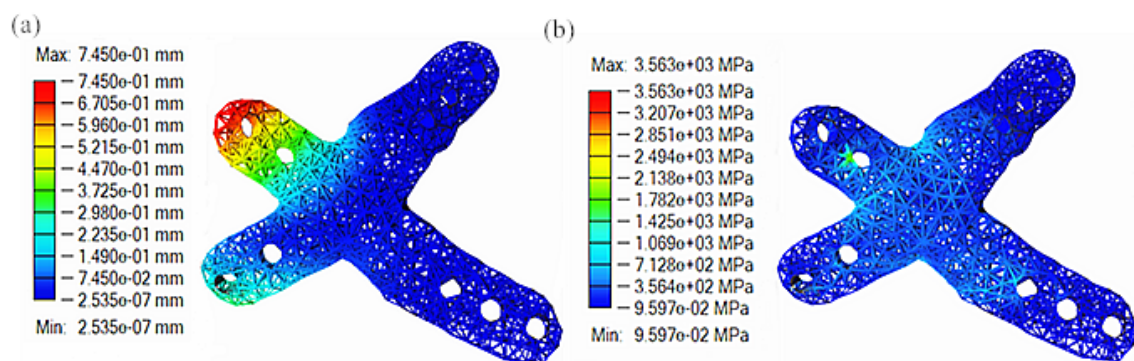


Figure 8. Fully porous plate based on topology optimization. (a) Displacement contour plot. (b) Stress contour plot.

(b) Field-driven fully porous plate

Finite element simulation results for the plate were exported from Inspire in Excel format, and the node positions of the model were further extracted from the file. A parametric script was developed to enable the parametric modeling of the porous biological fixation plate. Specifically, the Excel file containing node position data was imported into Rhinoceros via the Grasshopper plug-in to acquire the node information of the bone plate finite element

model. On this basis, a scalar plot was generated based on the node stress values derived from the explicit dynamic analysis, and a diamond-porous structural unit was constructed accordingly. Ultimately, all pores of the plate were transformed into diamond curved porous structures, with the structural distribution regulated by the stress field, as presented in Figure 9. This field-adaptive diamond-porous structure dynamically changes local structural parameters following the stress distribution of the pelvic

plate, forming an irregular gradient configuration. Such complex geometry poses enormous challenges for finite element meshing and numerical computation, rendering it impossible to obtain valid quantitative mechanical indicators, including displacement, stress distribution, safety factor, porosity, and equivalent elastic modulus, at the current research stage. Figure 9 shows that the porous structure density of the bio-fixed plate driven by the stress field increased at high-stress positions and decreased at low-stress positions, effectively improving the uneven stress distribution of the bio-fixed plate. Compared with traditional fully porous bone plates, the field-driven design optimizes the structural layout in response to stress variations, realizing a more homogeneous stress distribution and possessing better potential to reduce stress shielding. The field-driven biological fixed plate was similar to topology optimization and could effectively improve biocompatibility. However, the overall bearing capacity was reduced, so this design is particularly suited to pelvic non-major load-bearing sites and fracture areas where implants do not require secondary surgical removal, and it is not suitable for high-load primary force-bearing positions. This field-driven porous structure is regulated by the stress distribution to adaptively adjust strut thickness and improve stress uniformity. Due to the large model scale and high computational cost of the gradient porous

structure, detailed quantitative analyses and comparisons will be carried out in future research. This study mainly explores the feasibility of a field-driven structural design method.

3.5. Optimization of the 3D printing and manufacturing process for pelvic plates

3.5.1. Data processing of pelvic plates

When parts are printed in 3D, different placements and additional methods result in varying support addition amounts and molding layer thicknesses, directly affecting molding quality and part efficiency. Based on previous research findings on support addition and placement methods for 3D-printed parts, adjustments to the placement orientation of the pelvis and bone plate, as well as the supplementary arrangement of support structures, were implemented.²⁹ When printing the pelvis using a fused deposition modeling (FDM) printer, it can be positioned at a 15° incline relative to the baseplate to improve machining efficiency and avoid long scan lines. Considering the large machining section, the preheating temperature of the substrate was increased, and a 25% support was added to reduce stress concentrations and prevent warping. SLM molding technology was used to print plates. The plates were then imported into Magics for processing risk analysis. The analysis is shown in Figure

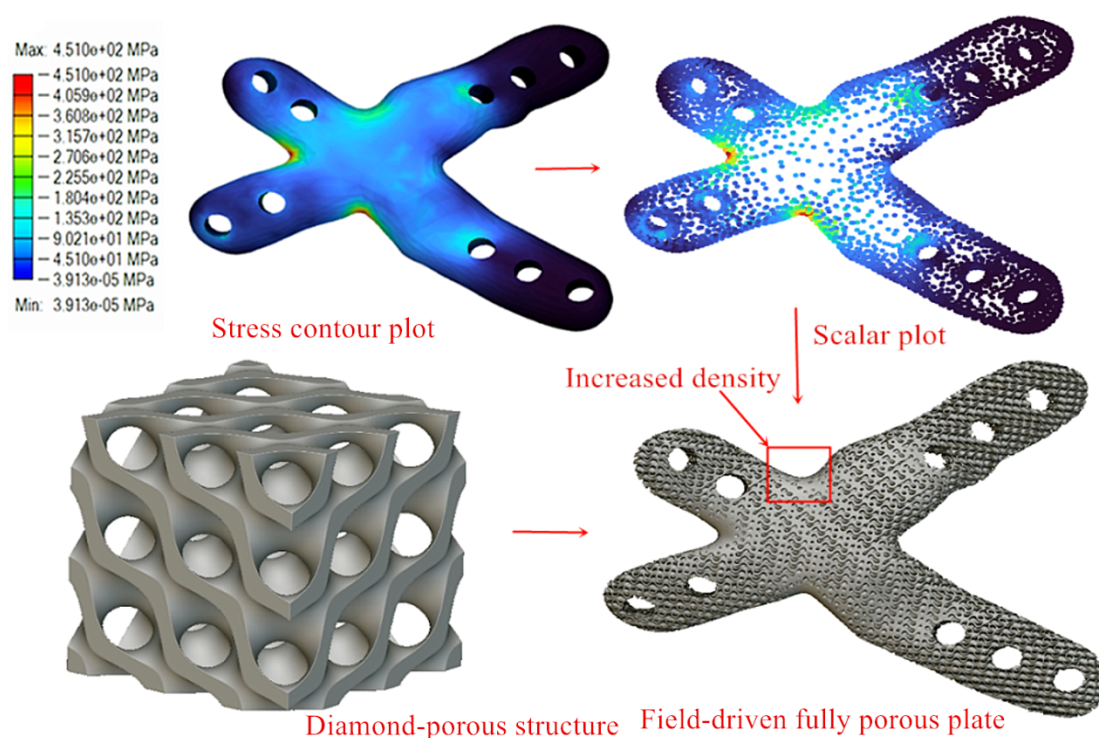


Figure 9. Full porosity simulation results of the biofixed bone plate

10a. The processing risks for the solid and partially porous plates at 60° of tilt were at the upper and lower ends (red and yellow regions indicate parts that require haptics). The inclination of 60° increased the height of the parts and the processing time. The processing risk profile was small, ensuring the surface quality of the plate as much as possible; by placing and optimizing all porous plates several times, the addition of support in the pores was minimized to ensure the molding quality when tilted approximately 50°. Similar to the processing risk analysis results, the solid and partially porous plates had less support addition and powder waste, as shown in Figure 10b. The fully porous plates provided slightly more support. To ensure molding quality, the solid and partially porous plates near the molded substrate partial support were encrypted.

3.5.2. Analysis of the forming effect of the pelvic bone plate

The SLM-printed pelvic plates are shown in Figure 11; they had a bright surface, a clear mesh structure, better metal texture, and less slag hanging. No evident warping deformation was found—there was an excellent overlap between porous structure struts. There was no macroscopically visible collapse, suggesting that SLM-shaped plates have excellent molding efficacy and surface quality. The structure between the porous structure struts of the partial porosity bone plate was clear; the struts were well lapped; the melt and the amount of slag hanging were small, as shown in Figure 11a. Although some splashing

particles were on the surface of the personalized bone plate, the melt was well lapped between them, which does not affect use after polishing. All porous-plated plates had clear structures between the struts and good lap plating (Figure 11b). There was a slightly higher amount of powder adhesion and sludging on the surface, particularly at the strut intersection, which may be related to the higher amount of overhang that occurs with SLM molding. However, this phenomenon does not affect use after processing (e.g., sandblasting, polishing).³⁰ SLM technology for Ti6Al4V is mature, and the dimensional deviation between printed parts and computer-aided design models is extremely small. These findings suggest that the SLM-printed pelvic bone plate meets use requirements.

3.5.3. Matching analysis of pelvic bone plates

The support structures of the 3D-printed pelvic bone plate were carefully removed using tools such as scissors, tweezers, knives, and cutters. SLM-printed personalized pelvic plates were assembled using FDM printing to assess fit, as shown in Figure 12; there was no apparent assembly conflict between the plate and the pelvis, and the fit was tight, demonstrating the individualization of the design. There was no noticeable assembly gap between the three plates and the pelvis; the fit was tight, and the screws and holes were in an appropriate position to meet the assembly requirements. Moreover, there was no significant interference between the plate and the pelvis, suggesting that personalized biological fixation plates designed

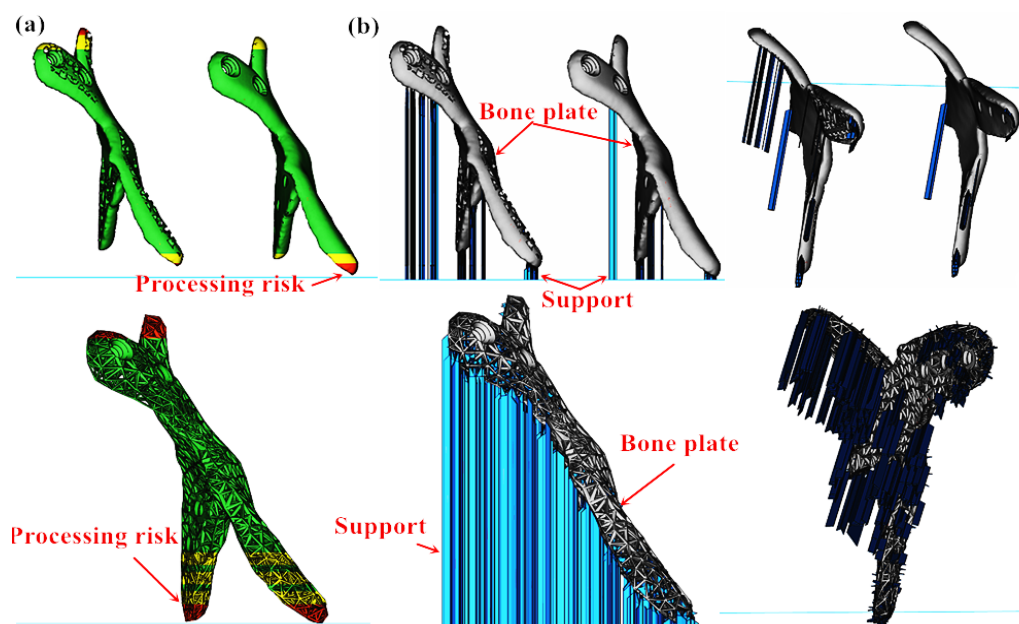


Figure 10. Data processing of a selective laser melting-printed bone plate. (a) Processing risk analysis. (b) Placement and support addition.

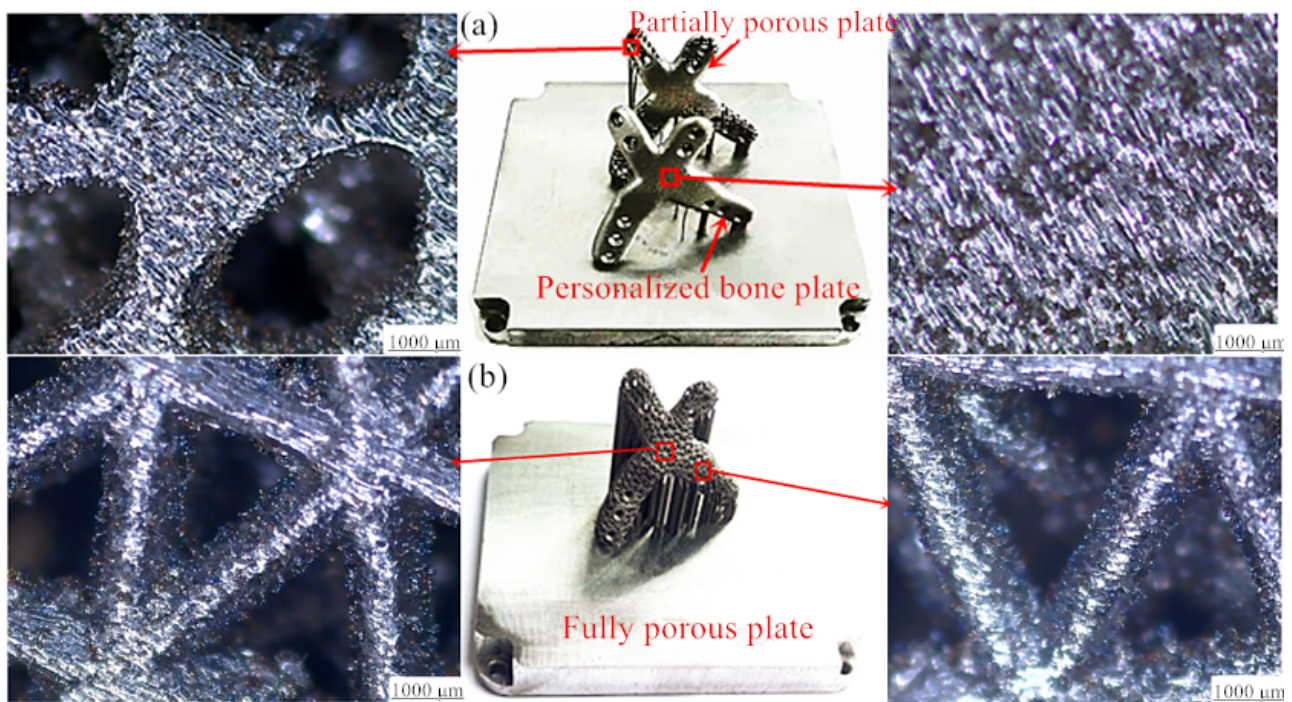


Figure 11. Selective laser melting-printed plate. (a) Personalized and partially porous plate. (b) Fully porous plate. Scale bars: 1,000 μm ; magnifications: 10 $\times$.

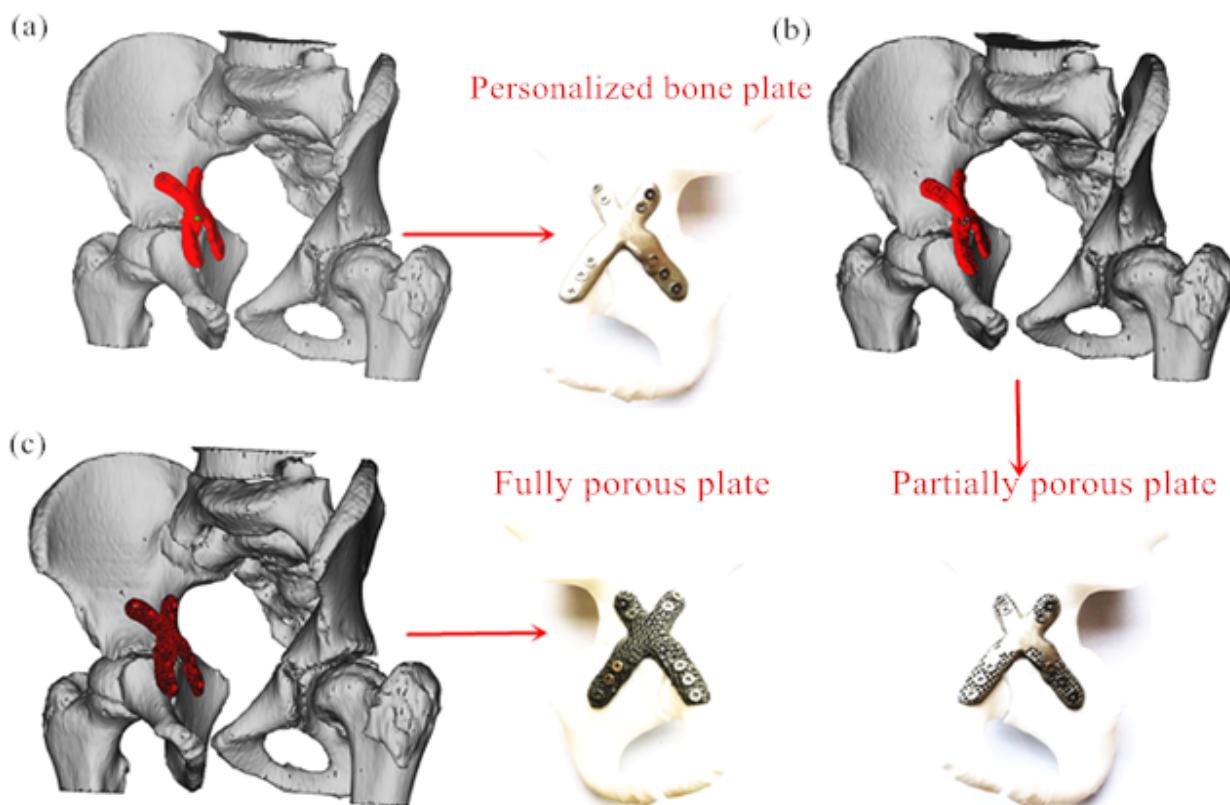


Figure 12. Matching test of plates. (a) Personalized plate. (b) Partially porous plate. (c) Fully porous plate.

based on 3D printing technology meet use requirements. This is only a qualitative fitting test, and quantitative gap measurement will be conducted in subsequent studies.

4. Conclusion

Several conclusions can be made from this study:

- (i) The deformation of the topology-optimized plate increased by approximately 15% compared with the original solid plate while remaining at a low level. The stress concentration increased slightly, reaching a maximum of 6.42×10^2 MPa, and no obvious local stress concentration points were observed. The final weight was reduced by 30.77%, successfully realizing the expected lightweight design target. These results are from a simplified finite element analysis and are intended for preliminary structural performance comparison.
- (ii) Finite element simulation of the partially porous plate was not performed due to the high computational cost. The partially porous structure was designed to achieve a trade-off between mechanical properties and biocompatibility, which will be verified in future studies. The plate's biocompatibility may be significantly improved with full porosity, which is more conducive to tissue ingrowth. However, the stress concentration and deformation increased. Therefore, careful consideration should be given to actual use. The above design and verification are supported by the existing mechanical and biocompatibility data on porous structures,²⁸ which meet the relevant design requirements for bone plates.
- (iii) The SLM-fabricated plates exhibited good surface quality, clear structural forming, no obvious warping, and tight fitting with the pelvic model. These results confirm the feasibility of the design and 3D-printing manufacturing process proposed in this study, providing a technical basis for the further preclinical development of high-performance, personalized pelvic bone plates.

The current study has limitations, including a small sample size (resulting in insufficient data representativeness), the adoption of computational simplifications (failing to reflect actual clinical conditions), and a lack of in vivo data (making it difficult to evaluate indicators such as degradation and osseointegration). To advance the practical implementation of 3D printing and bone plate technology, subsequent research should take the following measures: expand the sample size and design samples with porous structures mimicking natural bone; optimize the simulation model and verify it through experiments; conduct in vivo experiments on

large animals; and optimize the SLM process to ensure the porous structure promotes blood supply without risks, thus laying a foundation for clinical application.

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Conflict of interest

The authors report no conflicts of interest.

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Ethics approval and consent to participate

Not applicable.

Consent for publication

Not applicable.

Availability of data

The datasets used and/or analyzed during the current study are available from the corresponding author on reasonable request.

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