

REVIEW ARTICLE

Engineering science of additively manufactured patient-specific mandibular implants

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Abstract

Mandibular reconstruction is a major biomechanical challenge because it requires simultaneous restoration of anatomical form and structural stability under dynamic masticatory loading. While the structural degradation and poor conformity inherent to manually bent systems are overcome by patient-specific mandibular reconstruction plates (PSMRPs) fabricated via laser powder bed fusion (LPBF), clinical longevity remains constrained by manufacturing-induced defects and fatigue failure. In this critical review, the foundational engineering science governing LPBF-fabricated titanium alloy PSMRPs is examined, and an integrated paradigm spanning manufacturing science, computational biomechanics, structural optimization, and experimental validation is established. The process–structure–property relationships in LPBF are systematically evaluated, with specific detail given to how porosity morphology, residual stress, microstructural anisotropy, and surface integrity are dictated by volumetric energy density and complex thermal histories, ultimately governing fatigue crack initiation under cyclic loading. Moving from material to system-level mechanics, assembly failure modes are analyzed, while the biomechanical performance of locking versus non-locking fixation is compared. The role of finite element analysis alongside advanced structural optimization strategies is critically evaluated, and the mitigation of stress concentrations and stress shielding through size, shape, and topology optimization is demonstrated. Furthermore, state-of-the-art experimental validation methodologies are scrutinized, with emphasis placed on full-field strain characterization via digital image correlation and biomimetic cyclic fatigue testing. Finally, future translational frontiers are outlined, including artificial intelligence-driven generative design and mechanobiological simulations of long-term osseous remodeling. Collectively, the evidence suggests that the mastery of the interplay between additive manufacturing parameters and multi-scale biomechanics is paramount to the engineering of predictable, optimized cranio-maxillofacial patient-specific implants.

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1. Introduction

Mandibular reconstruction remains one of the most biomechanically demanding and technically challenging procedures in cranio-maxillofacial surgery because it requires simultaneous and precise restoration of facial aesthetics, mastication, speech, swallowing, occlusion, and long-term structural stability.¹⁻³ Segmental mandibular defects resulting from oncologic resection, trauma, osteoradionecrosis, or severe infection frequently necessitate reconstruction strategies capable of restoring both anatomical continuity and biomechanical functionality under highly dynamic loading conditions during daily masticatory activities.³

Although vascularized free flaps combined with fixation systems remain the clinical gold standard for large mandibular defects, fixation-related complications such as plate fracture, screw loosening, stress shielding, nonunion, and implant exposure persist as major contributors to postoperative morbidity and often require revision surgery.^{4,5} Conventional manual-bent mandibular reconstruction plates (MBMRPs) are additionally limited by poor anatomical conformity, intraoperative contouring variability, and fatigue weakening associated with repeated bending procedures.⁶ These inherent limitations regularly compromise fixation stability, increase the risk of long-term mechanical failure, and lead to suboptimal functional and aesthetic outcomes for patients.^{6,7}

The emergence of patient-specific mandibular reconstruction plates (PSMRPs), enabled by computer-aided design/computer-aided manufacturing technologies and additive manufacturing (AM), has transformed the reconstructive workflow by allowing implants to be designed directly from patient-specific anatomical data.⁸⁻¹⁰ Among available AM technologies, laser powder bed fusion (LPBF) has become the dominant approach for fabricating titanium-based cranio-maxillofacial implants because of its exceptional geometric freedom, high dimensional precision, and compatibility with medical-grade Ti6Al4V alloys.¹¹⁻¹³

Despite these advances, the long-term clinical reliability of additively manufactured mandibular implants remains constrained by unresolved engineering challenges. These involve process-induced defects, residual stress accumulation, poor surface integrity, anisotropic mechanical behavior, and fatigue-driven failure mechanisms.^{14,15} Consequently, successful clinical translation of PSMRPs requires integration of manufacturing science¹⁶⁻²⁰, computational biomechanics²¹⁻²³, structural optimization²⁴⁻²⁷, and experimental validation²⁸⁻³⁰ within a unified engineering framework.

Finite element analysis (FEA) and biomechanical experimentation have therefore emerged as indispensable tools for evaluating implant performance and optimizing fixation strategies.^{31,32} Computational biomechanics enables quantitative prediction of stress/strain distribution and implant deformation under physiological loading conditions^{24,33}, while experimental techniques such as cyclic fatigue testing and digital image correlation (DIC) provide essential validation of numerical simulations³⁴⁻³⁶. Simultaneously, emerging developments involving artificial intelligence (AI)-assisted implant design³⁷, mechanobiological simulation³⁸⁻⁴⁰, and automated digital workflows^{22,41} are beginning to reshape the future of patient-specific mandibular reconstruction.

Accordingly, this review critically examines the current engineering science underlying additively manufactured PSMRPs, with particular emphasis on LPBF process challenges and advancements, process-structure-property relationships, biomechanical optimization, computational simulation, experimental validation, and future translational directions. Unlike previous reviews focused separately on implant design, manufacturing processes, or computational biomechanics, this review uniquely integrates LPBF process-structure-property relationships with biomechanical optimization and experimental validation within a unified framework. By synthesizing these multidisciplinary perspectives, the review seeks to provide valuable insights that support the ongoing development of more reliable, durable, and biologically compatible patient-specific solutions for mandibular reconstruction.

To provide a clear overview of the interconnected factors discussed in this review, the key relationships between manufacturing factors, microstructural/defect features, mechanical consequences, clinical risks, and mitigation strategies are summarized in [Table 1](#).

2. Additive manufacturing of patient-specific mandibular reconstruction plates

2.1. Laser powder bed fusion advances patient-specific reconstruction

Additive manufacturing has fundamentally transformed the development of patient-specific mandibular reconstruction systems by enabling fabrication of implants with complex geometries, anatomical conformity, and individualized fixation strategies that cannot be achieved using conventional manufacturing.^{10,67-69} Among metal AM technologies, LPBF has emerged as the preferred fabrication method for craniomaxillofacial implants because of its high dimensional accuracy, superior geometric freedom,

Table 1. Summary of interconnected manufacturing, microstructural, mechanical, clinical, and mitigation aspects in additively manufactured patient-specific mandibular implants

Manufacturing factors	Microstructural/defect features	Mechanical consequences	Clinical risks	Mitigation strategies	References
Laser powder bed fusion process parameters (laser power, scanning speed, hatch spacing, layer thickness → volumetric energy density [VED])	Porosity: Lack of fusion (irregular, sharp) at low VED Keyhole porosity (spherical) at excessive VED	Stress concentration at pore edges Reduced effective load-bearing area Preferential fatigue crack initiation sites	Fatigue-driven implant fracture Premature implant failure under cyclic masticatory loading	VED optimization within a stable processing window <i>In situ</i> melt pool monitoring Hot isostatic pressing to close internal pores	42-47
Build orientation & thermal history (directional solidification, cooling rates)	Microstructural anisotropy: Columnar prior-β grains aligned with the build direction Non-equilibrium α' martensite	Orientation-dependent yield strength, ductility, and fatigue life Lower toughness when loaded perpendicular to layers	Unpredictable implant deformation Directional fatigue susceptibility in high-stress regions	Align the critical load-bearing axis with optimal build orientation Scan vector rotation (e.g., chessboard/island scanning)	47-52
Layer-wise thermal gradients	Residual stress: Tensile stress at surfaces Compressive stress in core	Part distortion, warping, and delamination from the build plate Superimposed tensile mean stress accelerates fatigue crack growth	Dimensional inaccuracy → poor anatomical fit Reduced effective fatigue strength → premature failure	Stress-relief heat treatment (post-process) Optimized scanning strategies (e.g., short scan vectors, pre-heated build plate)	53-58
Downward-facing surfaces (overhangs built against the powder bed)	High surface roughness: Adhered partially melted particles Staircase effect Surface-connected porosity	Severe stress concentration at surface asperities Early fatigue crack nucleation from surface/subsurface defects	Significantly reduced fatigue life Increased risk of implant fracture originating from the underside	Post-process surface finishing (machining, chemical polishing) Design for minimized down-skin regions or sacrificial supports	24,25,59-61
Implant design methodology (manual-bent vs. patient-specific)	Work hardening, microcracks, and residual stress from manual bending (MBMRP) As-built defect population in patient-specific mandibular reconstruction plates (PSMRPs)	MBMRP: embrittlement and low fatigue resistance PSMRP: superior geometric conformity, more homogeneous stress transfer	MBMRP: high risk of fatigue fracture and poor fit PSMRP: potential stress shielding if overly stiff	Use PSMRP to eliminate bending damage Apply structural (size/shape/topology) optimization to tailor stiffness and reduce stress	6-8,30,62,63
Fixation strategy (locking vs. non-locking system)	Locking: rigid screw-plate threaded interface Non-locking: friction-dependent plate-bone compression	Locking: high construct stiffness, resistance to screw pull-out, reduced micromotion Non-locking: requires perfect plate-bone adaptation; screws prone to toggling	Locking: risk of stress shielding and bone resorption under the implant Non-locking: risk of screw loosening, instability, and nonunion	Selective use of locking fixation in large defects, combined with stiffness-optimized implants Hybrid fixation strategies	21,23,64-66
Lack of integrated validation (reliance on simple static models)	(N/A: computational/experimental design gap)	Inaccurate prediction of dynamic, patient-specific fatigue behavior and bone remodeling	Unvalidated implant designs entering clinical use Unexpected long-term mechanical or biological complications	Integrate finite element analysis, cyclic fatigue testing with digital image correlation, and mechanobiological simulation into a unified development workflow	22,26,30,32,41

and compatibility with titanium alloys.^{11,70} In particular, LPBF enables fabrication of thin-walled structures, curved fixation regions, and patient-matched screw trajectories that are difficult or impossible to manufacture using subtractive machining or manual bending techniques, thereby substantially improving anatomical adaptation and surgical reproducibility.^{8,63}

In LPBF, thin layers of metallic powder are selectively melted by a high-energy laser according to a predefined digital model generated from patient imaging data.¹¹ Layer-wise fabrication allows direct production of highly complex patient-specific implants with minimal material waste and substantial design flexibility.⁷¹ This manufacturing paradigm also enables rapid conversion of patient computed tomography (CT) data into customized implants without intermediate tooling or mold fabrication, thereby shortening production cycles and facilitating efficient personalization in complex craniomaxillofacial reconstruction.^{25,30} Consequently, LPBF enables fabrication of reconstruction plates specifically adapted to mandibular anatomy, thereby improving implant fit, reducing intraoperative contouring, and minimizing surgical variability.

Compared with conventional MBMRPs, LPBF-fabricated PSMRPs demonstrate improved geometric precision and more homogeneous stress distribution across the reconstruction assembly.^{7,30} Several finite element and experimental studies have shown that patient-specific plates can reduce stress concentration around osteotomy sites and fixation screws by improving load transfer continuity and structural conformity, which may contribute to reduced risks of plate fracture and screw loosening.^{21,23,24} Moreover, AM facilitates integration of advanced design strategies, including topology optimization^{26,27,72–76}, lattice structures^{22,29}, and stiffness-tailored geometries^{24,25} aimed at improving biomechanical performance and reducing implant weight.

Recent developments in AM increasingly emphasize patient-specific reconstruction not only as a geometric problem but also as a biomechanical and biological optimization challenge. This shift requires integrating mechanobiological simulation models to match implant stiffness to the mechanical properties of cortical or cancellous bone, thereby reducing stress-shielding and actively stimulating bone graft regeneration.^{39,77} Consequently, manufacturing strategies are progressively shifting toward fatigue-informed implant design and personalized biomechanical adaptation.

2.2. Process–structure–property relationships

The mechanical performance and long-term reliability

of LPBF-fabricated mandibular reconstruction plates are strongly governed by process–structure–property relationships established during AM.^{11,78} This dependency arises because the laser-induced thermal cycles directly control the density and morphology of defects, the microstructural phase constitution, and the internal stress state, all of which determine the implant's response to physiological loading.^{13,79} Rapid melting and solidification cycles during LPBF generate complex thermal histories that directly influence microstructural evolution, residual stress accumulation, porosity formation, and anisotropic mechanical behavior.^{70,80} Specifically, high cooling rates induce a non-equilibrium phase transformation from the equilibrium $\alpha+\beta$ microstructure to a brittle, needle-like α' martensitic microstructure, which significantly restricts dislocation movement and increases localized internal stresses.^{13,14,81}

Volumetric energy density (VED), determined by laser power, scanning speed, hatch spacing, and layer thickness, represents one of the most influential processing variables controlling implant quality.^{49,82} A stable processing window for LPBF of Ti6Al4V is generally achieved at volumetric energy densities between 60 and 70 J/mm³, within which superior relative density, surface roughness, and microhardness are obtained concurrently.^{49,83} Insufficient energy input may result in lack-of-fusion defects and incomplete densification, whereas excessive energy density may induce keyhole porosity, thermal instability, and dimensional distortion.⁸³ This occurs because inadequate VED limits the overlap between adjacent melt pools, leaving unmelted powder tracks, whereas excessive VED vaporizes volatile alloying elements such as aluminum, causing melt-pool instability and collapse, which can trap spherical gas pores and keyhole-induced voids within the solidified material matrix.^{43,82} Previous studies have also demonstrated that lack-of-fusion pores usually exhibit irregular geometries with sharp stress concentrations⁴⁴, while keyhole pores are typically spherical and associated with vaporization instability⁴⁶, both of which markedly compromise fatigue resistance and long-term biomechanical reliability⁴⁷.

Recent literature increasingly recognizes that implant performance cannot be adequately evaluated based solely on static strength.³⁵ Instead, fatigue behavior, crack propagation resistance, and defect morphology have emerged as dominant determinants of clinical implant survival.^{14,35} Surface-connected porosity and manufacturing-induced irregularities may act as preferential crack initiation sites under cyclic mastication loading.^{14,42,45} These defects promote early fatigue crack nucleation through localized stress amplification, with

studies showing that surface and near-surface pores are particularly detrimental compared to fully internal voids.^{42,45} The risk is particularly pronounced near screw holes, where geometric stress concentration factors of 2–3 combine with near-surface defects and tensile residual stresses of up to 800–1000 MPa, creating a critical locus for crack initiation under the cyclic loads generated during mastication.^{55,56} This is particularly important for mandibular reconstruction plates because the oral maxillofacial region is continuously subjected to repetitive chewing forces, where small surface or subsurface defects can progressively evolve into fatigue cracks and eventually lead to catastrophic implant fracture.

Another critical consideration involves anisotropic microstructure formation. Directional solidification during LPBF fabrication produces orientation-dependent mechanical properties that may influence implant stiffness⁸⁴, ductility⁵², and fatigue life⁸⁵. This anisotropy stems from the preferential growth of columnar prior- β grains aligned with the build direction parallel to the maximum thermal gradient, which creates distinct microstructural slip planes that result in lower yield strength and fracture toughness when loads are applied perpendicular to the layer boundaries.^{48,51,81} Consequently, optimization of LPBF process parameters must simultaneously address manufacturability, microstructural integrity, and biomechanical performance. A holistic approach integrating parameter optimization, post-processing, and build orientation considerations is required to ensure reliable clinical outcomes for patient-specific mandibular reconstruction plates.

2.3. Residual stress, surface integrity, and fatigue

Residual stress accumulation remains one of the principal limitations of LPBF-fabricated titanium implants.^{55,56,85} During layer-wise manufacturing, localized thermal gradients generate repeated expansion and contraction cycles that induce complex residual stress fields within the printed structure.⁵³ Specifically, the rapid cooling of newly deposited upper layers induces a tensile stress state at the component's surface balanced by compressive stresses in the core, which can exceed the yield strength of the titanium alloy and cause severe delamination from the build plate or geometric warping.⁵⁵ These stresses may lead to distortion, dimensional inaccuracy, reduced fatigue resistance, and premature crack propagation.⁵⁶ The magnitude and distribution of residual stress are strongly influenced by scanning strategy, build orientation, and thermal accumulation effects, and excessive tensile residual stress may even cause part delamination or cracking during fabrication of thin-walled titanium structures.⁵⁷

Fatigue failure is particularly relevant in mandibular reconstruction because fixation plates are exposed to repetitive cyclic loading generated during mastication and speech.^{58,86} Unlike acute overload failure, fatigue fracture develops progressively through crack initiation and propagation under repetitive subcritical loading conditions.^{52,58,87} Clinical studies increasingly identify fatigue-related plate fracture as a major cause of long-term reconstruction failure.^{14,88} Biomechanical analyses have shown that mandibular reconstruction plates may experience millions of loading cycles annually, with stress concentrations frequently localized near screw holes and mandibular defect regions, thereby substantially increasing the risk of fatigue crack initiation under long-term service conditions.^{6,14,30,52,58,86}

Surface integrity plays a crucial role in fatigue performance because surface roughness and partially fused particles act as stress concentration sites.^{59,61} Downward-facing surfaces fabricated against partially sintered powder generally exhibit inferior surface quality compared with upward-facing regions, thereby increasing susceptibility to crack initiation.^{50,60} This structural disparity arises because the underlying loose powder bed has a thermal conductivity an order of magnitude lower than the solid metal substrate, causing a larger, unstable melt pool that sags and traps semi-molten satellite particles along the overhanging down-skin boundaries.^{79,89} Studies on LPBF of Ti6Al4V have also demonstrated that surface asperities, adhered particles, and staircase effects significantly reduce fatigue strength by promoting localized stress accumulation and accelerating early-stage crack formation.^{15,81,90} For example, as-built LPBF-fabricated Ti6Al4V typically exhibits surface roughness (Ra) of 10–25 μm , with downward-facing surfaces exceeding 20 μm , which can reduce the fatigue limit at 10^7 cycles by 40–70% relative to machined or polished counterparts.^{60,61}

Several engineering strategies have therefore been proposed to improve fatigue resistance of LPBF-fabricated implants.^{91,92} Heat treatment and hot isostatic pressing are commonly employed to reduce residual stress and internal porosity, while optimized scanning strategies such as chessboard hatching and rotating scan vectors may reduce thermal gradients and distortion.^{54,92,93} In particular, hot isostatic pressing has been reported to effectively close internal pores and reduce defect populations⁹³, thereby improving the fatigue life of LPBF-fabricated titanium alloys⁵⁴, whereas scan vector rotation can redistribute thermal accumulation and mitigate anisotropic residual stress development^{53,57}.

Current research increasingly emphasizes fatigue-informed implant design as a critical requirement for

achieving long-term biomechanical reliability in patient-specific mandibular reconstruction. Accordingly, recent engineering approaches integrate finite element fatigue prediction, topology optimization, and defect-tolerant design principles to improve load transfer behavior and extend the functional lifespan of patient-specific mandibular implants under physiological cyclic loading.

3. Biomechanical optimization of patient-specific mandibular reconstruction plates

3.1. Failure mechanisms in reconstruction assembly

Mechanical complications remain among the most significant limitations of mandibular reconstruction systems.^{1,3,9} Despite advances in surgical planning and implant manufacturing, reconstruction plates continue to experience complications, including fatigue fracture, screw loosening, stress shielding, nonunion, and plate exposure.^{5,14,21} Clinical follow-up studies have reported that these complications are strongly associated with complex mandibular loading environments, large segmental defects, and long-term cyclic functional loading, which collectively challenge the mechanical durability of reconstruction assemblies.^{22,24,63} It is also indicated that these complications often require revision surgery and compromise long-term functional outcomes.^{1,4} General fatigue mechanisms^{14,25,26} of the PSMRP are illustrated in [Figure 1](#).

Fatigue fracture represents one of the most clinically relevant failure mechanisms because mandibular reconstruction plates are continuously subjected to repetitive loading during chewing, swallowing, and speech.^{6,14} Finite element and retrieval analyses have demonstrated that stress concentration commonly localizes near screw-plate interfaces and plate curvature transitions, where repeated tensile-compressive stress cycling accelerates crack nucleation and fatigue propagation.^{24,30,94} These stress risers can elevate local stresses to several times the nominal applied load, driving crack initiation and propagation under typical masticatory forces over hundreds of thousands of cycles.

Stress shielding arises when the high-modulus titanium alloy plate bears the majority of the load, reducing strain in the underlying bone below the threshold required for Wolff's law-mediated remodeling, often leading to bone resorption within weeks to months.^{23,73,95} Excessive construct stiffness may additionally contribute to stress shielding and altered physiological load transfer, potentially impairing bone remodeling and osseointegration.^{27,74} Meanwhile, insufficient fixation stability may result in micromotion, delayed union, and screw loosening.^{21,23} This biomechanical mismatch is particularly important in titanium reconstruction systems because the elastic

modulus of Ti6Al4V substantially exceeds that of cortical bone, thereby reducing physiological stress stimulation required for normal bone remodeling and healing.

These challenges highlight the need for biomechanically optimized reconstruction systems capable of balancing structural rigidity with physiological flexibility while minimizing fatigue accumulation and stress concentration.⁹⁶ Accordingly, recent implant design strategies increasingly incorporate topology optimization^{26,27,75}, porous architectures^{22,29,97}, and functionally graded structures⁹⁸ to achieve improved load transfer behavior and enhanced long-term biomechanical compatibility.

3.2. Finite element analysis and computational biomechanics

Finite element analysis has become an indispensable tool for evaluating the biomechanical behavior of patient-specific mandibular reconstruction systems.^{31,32} FEA enables numerical simulation of stress distribution, strain transfer, fixation stability, and implant deformation under physiologically relevant loading conditions.^{10,99} By integrating patient-specific anatomical geometry derived from CT imaging with material constitutive models, FEA can quantitatively predict mechanical responses of mandibular reconstruction assemblies before surgical implementation.^{30,75}

Compared with experimental testing alone, computational biomechanics allows rapid design iteration^{25,62} and assessment of multiple fixation strategies^{21,23} without invasive experimentation. Consequently, FEA is widely employed during preclinical optimization of reconstruction plates to identify mechanically vulnerable regions^{24,62} and evaluate the influence of implant geometry²⁶, screw configuration⁶, and loading conditions³⁰. This numerical predictive capability drastically compresses the research-and-development loop by enabling automated parametric variations of plate topology²⁴⁻²⁶, screw arrangements^{6,62}, and fixation schemes^{21,23} to be screened and discarded before entering physical production, substantially reducing experimental cost and development time.

However, important limitations remain in current mandibular FEA studies.^{31,32} Many computational models continue to rely on simplified isotropic bone assumptions and static loading conditions that inadequately reproduce the heterogeneous and dynamic biomechanics of the reconstructed mandible.^{22,23,25} Furthermore, muscle force simulation often lacks physiological realism.^{38,62} In reality, mandibular biomechanics involve anisotropic bone behavior^{29,100}, temporally varying occlusal loading^{24,26,30}, and coordinated activation of multiple masticatory

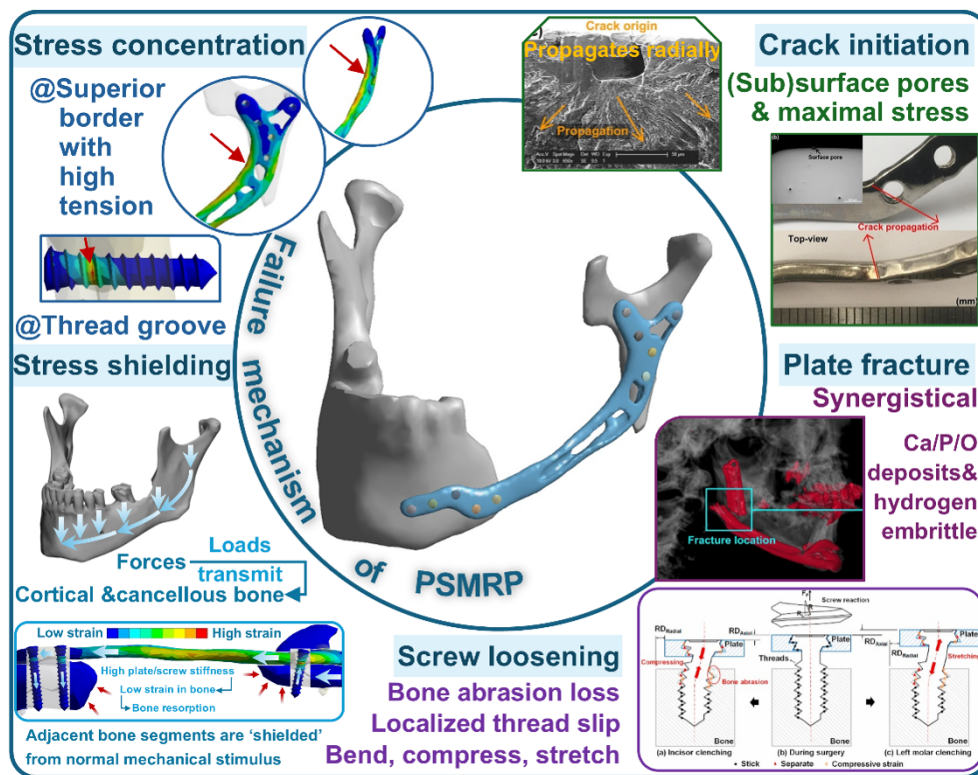


Figure 1. General failure mechanisms of patient-specific mandibular reconstruction plates

muscles^{21,23-25}, all of which may significantly influence local stress distribution and fixation stability. The predictive reliability of mandibular FEA models is strongly influenced by key simplifying assumptions. Assigning homogeneous isotropic properties to cortical and cancellous bone, instead of regionally varying elastic moduli of 10–20 GPa and 0.1–2 GPa, respectively, using idealized temporomandibular joint constraints, and applying generic muscle loading vectors often lead to inaccurate stress and strain distributions.^{101,102} Additionally, modeling screw–bone and bone–plate interfaces as perfectly bonded or frictionless contacts tends to underestimate micromotion and localized stress concentrations, resulting in overly optimistic assessments of fixation stability and fatigue resistance.¹⁰²⁻¹⁰⁴

Recent developments increasingly combine patient-specific imaging, optimization algorithms, and computational biomechanics to establish integrated digital reconstruction workflows capable of predicting implant performance and postoperative adaptation.^{26,41} Emerging digital workflows integrating AI-assisted design^{37,105}, topology optimization^{26,27}, and biomechanical simulation^{6,22,38} have shown substantial potential for improving surgical precision, implant longevity, and personalized functional reconstruction outcomes.

3.3. Structural optimization strategies

Structural optimization has emerged as a powerful engineering approach for improving the biomechanical performance of PSMRPs.⁹⁶ It systematically modifies the plate's internal structure and external shape to better withstand the complex chewing forces transferred through the mandible.^{26,27} Size and shape optimization strategies aim to reduce stress concentration by modifying local geometric features, including thickness distribution¹⁰⁶, curvature²⁴, and transition regions⁷⁶. Such geometric refinement is particularly important in mandibular reconstruction because abrupt sectional variation and sharp curvature discontinuities may generate localized stress concentration that accelerates fatigue crack initiation under cyclic mastication loading.^{14,24}

Recent studies demonstrate that smooth tangent transitions in highly stressed mandibular angle regions may substantially reduce peak tensile stress²⁴ and delay fatigue crack initiation¹⁴. Practically, this means replacing sharp corners or stepped thickness changes with continuous, curved profiles that eliminate local stress risers. In addition, topology optimization enables algorithm-driven redistribution of material according to loading conditions, thereby generating lightweight structures with improved

structural efficiency.¹⁰⁷ By minimizing mechanically inefficient material while preserving critical load-bearing pathways, topology optimization can improve stiffness-to-weight ratio and reduce localized stress concentration in patient-specific fixation systems.⁷³

Topology-optimized reconstruction plates have demonstrated reduced implant weight, more homogeneous stress distribution, and improved biomechanical efficiency compared with conventional geometries.^{26,27,72} For example, optimized designs can reduce peak von Mises stress while simultaneously lowering implant mass relative to standard reconstruction plates.¹⁰⁸ However, optimization strategies must additionally consider AM constraints, including unsupported overhangs, minimum feature size, and powder removal accessibility.⁹⁶ Specifically, if design elements feature downward-facing surface angles below 45° relative to the build plate, they will fail during LPBF processing due to thermal warping unless temporary, sacrificial support structures are added.⁶⁰ Without incorporation of manufacturability constraints, optimized structures may contain geometries that are difficult to fabricate reliably using LPBF, potentially resulting in unsupported regions, incomplete powder evacuation, or reduced dimensional accuracy.^{60,96}

Future optimization workflows are expected to integrate fatigue prediction^{86,87}, manufacturability constraints⁹⁶, and mechanobiological criteria¹⁰⁹ into unified multi-objective frameworks for patient-specific implant generation. These next-generation computational frameworks increasingly combine FEA, biological remodeling algorithms, and AM process simulation to simultaneously optimize structural safety, fatigue durability, and bone regeneration potential.

3.4. Locking versus non-locking fixation

Fixation strategy strongly influences the biomechanical behavior of mandibular reconstruction systems. The choice determines how loads are transferred from the plate to the underlying bone, which directly affects healing and long-term stability.^{64,110} Conventional non-locking fixation relies on compression between the plate and bone generated by screw insertion torque. This means that friction at the plate–bone interface resists micromotion, requiring the plate to be perfectly contoured to the bone surface to achieve adequate stability.²³ This mechanism requires intimate plate adaptation and may compromise periosteal vascularization. Excessive plate–bone compression may impair local blood perfusion and increase the risk of cortical bone resorption, particularly in long-span mandibular reconstruction, where soft tissue vascularity is already compromised.⁵

In contrast, locking fixation systems establish a rigid

screw–plate interface in which stability is achieved through mechanical locking between screw heads and threaded plate holes.⁶⁵ This configuration eradicates dependence on plate–bone compression and may preserve local blood supply.¹¹¹ The threaded interface effectively transforms the plate–screw assembly into a monolithic framework, meaning the construct can withstand high shear and bending forces through structural rigidity alone, leaving a micro-gap that preserves the underlying cortical blood flow.²¹ Locking systems are generally less sensitive to imperfect plate adaptation and may improve fixation reliability in complex mandibular defects.

Biomechanical studies generally demonstrate superior fixation stability and reduced screw loosening in locking systems compared with non-locking configurations.^{21,110} Specifically, locking screws experience lower axial and shear micromotion under cyclic loading because the threaded interface prevents the toggling and back-out commonly observed in non-locking designs.⁶⁵ Nevertheless, increased construct stiffness may exacerbate stress shielding and alter physiological load transfer.^{66,111} Several finite element and experimental investigations have reported that locking plates reduce interfragmentary motion and screw pull-out risk⁶⁵, but excessively rigid fixation may simultaneously decrease physiological strain stimulation required for bone remodeling⁶⁶.

Although AM of threaded locking geometries remains technically challenging, ongoing improvements in LPBF precision and post-processing are expected to facilitate broader implementation of patient-specific locking reconstruction systems.¹¹² Recent advances in high-resolution LPBF^{89,113}, thread-finishing techniques⁵⁹, and precision machining^{8,10} have substantially improved dimensional accuracy and thread integrity of additively manufactured titanium fixation components.

3.5. Biomechanical superiority of patient-specific mandibular reconstruction plates

Patient-specific mandibular reconstruction plates consistently demonstrate superior biomechanical behavior compared with manually bent systems.^{6,7} Unlike MBMRPs, which require extensive intraoperative contouring, PSMRPs are designed directly from patient-specific anatomy and therefore provide improved geometric conformity and more homogeneous load transfer.^{8,63} Improved anatomical adaptation reduces unintended plate deformation and minimizes localized gap formation between the implant and bone surface, thereby promoting more favorable stress transfer across the reconstruction assembly.^{6,30}

Repeated bending procedures associated with MBMRPs may introduce residual stress accumulation,

work hardening, and microcrack formation that compromise fatigue resistance.^{14,30,58} Each bending cycle during contouring induces plastic deformation within the titanium microstructure, increasing dislocation density and brittleness while potentially nucleating microscopic cracks that propagate under cyclic masticatory loads.^{35,52} In contrast, PSMRPs eliminate many of these mechanically detrimental processes and consequently exhibit improved long-term durability.

Experimental and computational investigations increasingly demonstrate that PSMRPs provide reduced peak stress concentration, improved fixation stability, higher stiffness, and prolonged fatigue life compared with manually bent reconstruction systems.^{6,30} For instance, studies have shown that PSMRPs can withstand over 5 million fatigue cycles at physiological loads, whereas MBMRPs frequently fail before 2 million cycles under identical conditions.³⁰ Moreover, improved anatomical fit may reduce surgical time and enhance postoperative functional outcomes. Validated studies show PSMRPs can achieve over 30% higher stiffness and significantly extended fatigue life relative to MBMRPs under physiological loading.³⁰

The growing integration of computational biomechanics with experimental validation further strengthens evidence supporting the biomechanical superiority of patient-specific mandibular reconstruction systems.^{27,30,41} Finite element analyses can now accurately predict plate performance under personalized loading conditions, and these predictions have been consistently validated by mechanical testing of additively manufactured prototypes^{7,22,27,30}, thereby improving confidence in the translational reliability of patient-specific mandibular reconstruction strategies.

4. Experimental validation

4.1. Cyclic fatigue testing

Experimental validation remains essential for evaluating the long-term reliability of mandibular reconstruction systems. Biomechanical testing under controlled load conditions is required to confirm whether a plate design can withstand months or years of repeated physiological loading before clinical translation or regulatory evaluation.^{114,115} Among available biomechanical approaches, cyclic fatigue testing has emerged as the most clinically relevant method because mandibular reconstruction plates are continuously exposed to repetitive loading during mastication and speech.^{7,58} Physiological mastication may generate hundreds of thousands to millions of loading cycles annually, making fatigue behavior a dominant factor governing long-term

implant survival and structural durability.

Unlike quasi-static testing, which evaluates immediate structural strength, cyclic fatigue testing enables investigation of crack initiation, fatigue accumulation, and long-term implant survival.¹¹⁶ Quasi-static testing simply loads a specimen to failure in a single monotonic loading event, whereas cyclic testing, a process much closer to real-world implant degradation, reveals how microscopic damage gradually develops over time.¹¹⁷ Experimental studies consistently demonstrate that fatigue failure frequently initiates in regions characterized by stress concentration and manufacturing-induced defects.^{25,58} In LPBF-fabricated Ti6Al4V plates, fatigue cracks commonly nucleate at surface irregularities, internal porosity, and geometric discontinuities near screw holes or plate curvature transitions during repeated loading cycles, with as-built specimens showing markedly lower fatigue strength than post-processed specimens.

Recent research increasingly emphasizes physiologically realistic testing configurations incorporating dynamic muscle simulation, asymmetrical loading, and temporomandibular joint constraints.¹¹⁸ For example, instead of applying simple vertical loads to an isolated plate¹¹⁷, advanced test setups simulate the pull of masseter and temporalis muscles¹¹⁸ and reproduce the uneven bilateral forces that occur during unilateral chewing^{27,30}. Such approaches may substantially improve translational relevance and strengthen the correlation between computational prediction and clinical performance. By reproducing more realistic occlusal force vectors and mandibular kinematics, advanced experimental platforms can better replicate in vivo biomechanical environments and improve the predictive accuracy of implant performance assessment.

4.2. Digital image correlation

Digital image correlation has emerged as a powerful non-contact optical technique for evaluating full-field deformation and strain distribution in biomechanical systems.³⁶ DIC works by tracking the movement of random speckle patterns applied to the specimen surface across sequential images captured during loading, generating a continuous strain map over the entire plate surface.^{119,120} Unlike conventional strain gauges, DIC enables comprehensive assessment of local strain accumulation and implant deformation under dynamic loading conditions.^{121,122} Because DIC provides high-resolution full-field strain mapping without physical sensor attachment, it is particularly advantageous for analyzing complex geometries and heterogeneous deformation patterns in mandibular reconstruction assemblies.^{30,41}

In mandibular reconstruction research, DIC is increasingly used to validate finite element predictions and evaluate load transfer behavior in patient-specific reconstruction systems.^{22,30} Recent studies combining cyclic fatigue testing and DIC have demonstrated strong agreement between experimental strain measurements and computational simulations. High correlation coefficients between DIC-derived strain fields and finite element predictions have been reported for optimized reconstruction plates tested under masticatory loading^{30,41}. This combined experimental-computational approach enables quantitative comparison of strain distribution patterns, thereby improving confidence in numerical predictions of implant biomechanics and fixation stability.

The integration of DIC with advanced imaging technologies may further improve the biomechanical characterization of mandibular reconstruction systems under physiologically realistic loading environments. Recent developments integrating DIC with CT¹²³, high-speed imaging¹²⁴, and synchronized mechanical testing systems¹²⁵ have enabled multi-scale characterization of deformation behavior in complex biological structures. Combining DIC with micro-CT¹²⁶, for example, allows simultaneous tracking of surface deformation and internal bone-implant micromotion, offering a complete mechanical assessment that neither method alone can provide.

4.3. Validation of computational models

Despite widespread use of FEA in mandibular reconstruction research, insufficient experimental validation remains a major barrier to clinical translation.^{101,102} Without experimental confirmation, there is no guarantee that a finite element model's predictions accurately reflect the behavior of the actual physical implant when surgically placed.^{127,128} Many current validation approaches rely on simplified synthetic mandibular models that inadequately reproduce the anisotropic and heterogeneous properties of native bone.^{6,7,22,24,41} Synthetic polyurethane models often assume homogeneous isotropic properties^{28,100,114}, leading to discrepancies in predicted stress/strain distributions compared to real bone with regionally varying cortical thickness and trabecular architecture.

Furthermore, variability in loading configuration, muscle simulation, material properties, and boundary conditions complicates comparison between studies and limits reproducibility.^{102,103,129} Differences in bite force magnitude, muscle attachment modeling, and fixation boundary assumptions may substantially alter predicted stress distributions and deformation behavior across independent computational studies.¹⁰² For example, loads

may be applied directly to the plate surface^{30,41}, through a soft-tissue-simulating silicone pad¹¹⁸, or using a full cadaveric mandible^{22,104}; these methodological differences can yield different stress distributions and limit direct comparisons across studies.

Future validation strategies should therefore prioritize development of biomimetic mandibular models^{27,29}, physiologically realistic loading systems¹¹⁸, and standardized experimental protocols^{30,41}. Biomimetic models that replicate mandibular cortical thickness, trabecular architecture, and anisotropic mechanical properties using multi-material AM would substantially improve validation fidelity compared with homogeneous synthetic blocks.^{17,89,113,130} Integration of experimental testing with patient-specific computational simulation may ultimately establish more reliable predictive frameworks for personalized mandibular reconstruction.^{30,41} This approach—using the same patient's CT data to build both the finite element model and the additively manufactured experimental test specimen—ensures that validation directly assesses the clinical scenario rather than a generic approximation.

5. Future perspectives

5.1. Artificial intelligence-assisted implant design

Artificial intelligence is increasingly reshaping the development of patient-specific mandibular reconstruction systems. AI algorithms can now learn patterns from large datasets of patient images¹³¹ and implant designs¹³² to automate or augment tasks that previously required hours of manual work by engineers and surgeons. Machine-learning algorithms are beginning to support automated anatomical segmentation¹³³, topology optimization¹³⁴, implant generation¹³⁵, and prediction of biomechanical performance^{86,136}. Deep learning-based image analysis and generative design algorithms have demonstrated substantial potential for improving segmentation accuracy^{105,131}, accelerating implant customization^{137,138}, and reducing operator-dependent variability during digital surgical planning^{139,140}.

Artificial intelligence-assisted workflows may substantially reduce preoperative planning time while improving reproducibility and personalization.^{138,139} Future platforms are expected to integrate imaging, biomechanics, and AM into fully automated design ecosystems capable of generating implants optimized simultaneously for anatomical conformity, structural integrity, and fatigue resistance.^{139,140} Such integrated frameworks may enable rapid iterative optimization by coupling patient-specific imaging data with biomechanical simulation and manufacturability analysis, thereby facilitating more efficient and data-driven implant development.

5.2. Mechanobiological simulation

Most currently available finite element studies evaluate immediate postoperative biomechanics while neglecting long-term tissue healing and bone remodeling.^{32,33,87} This limitation is significant because postoperative mandibular adaptation involves continuous interactions among mechanical loading, tissue differentiation, vascularization, and bone remodeling throughout the healing process.^{141,142} Mechanobiological models incorporate algorithms such as strain-adaptive bone remodeling to simulate dynamic changes in bone density and microarchitecture over months of healing.^{40,143} Consequently, mechanobiological simulation has emerged as a major frontier in computational mandibular reconstruction research.

Mechanobiological models aim to predict how local mechanical stimuli influence tissue differentiation, bone regeneration, and fixation stability throughout healing.^{142,143} Specifically, these models evaluate implant load transfer during healing by assigning stage-dependent mechanical properties to granulation tissue, immature bone, and mature bone³⁸⁻⁴⁰, while using mechanobiological thresholds of strain, fluid flow, or octahedral shear stress to predict tissue differentiation into bone, cartilage, or fibrous tissue and thereby forecast union or nonunion outcomes^{38,39,143}. Integration of healing-stage simulation into mandibular biomechanics may substantially improve prediction of long-term implant performance and postoperative adaptation.^{38,74} For example, a plate that shows acceptable stress immediately postoperatively might later cause stress-shielding once the bone regains stiffness, leading to resorption; mechanobiological simulation can flag such delayed failures that static analyses miss.

Future computational frameworks are expected to combine mechanobiology^{38,39}, fatigue analysis^{86,87}, and patient-specific loading conditions^{24,26} into unified predictive models capable of simulating dynamic postoperative adaptation. By integrating time-dependent healing simulation with cyclic loading analysis and patient-specific functional biomechanics, future predictive models may provide more accurate estimation of implant longevity, bone adaptation, and risk of fixation failure.

5.3. Automated digital workflow

The current workflow for patient-specific mandibular reconstruction remains labor-intensive and fragmented, requiring substantial manual interaction during segmentation, implant design, finite element modeling, and surgical planning.^{24,25} In practice, a single case may require a radiologist to segment anatomy, an engineer to design the plate, a simulation specialist to run FEA, and a surgeon to plan the approach. This multi-step workflow

often involves repeated data conversion and software transitions, which may increase processing time, introduce user-dependent variability, and limit clinical efficiency.

Recent advances in AI-assisted segmentation, generative design, cloud-based surgical planning, and automated FEA are beginning to accelerate implant development while improving reproducibility and standardization.^{138,139} For example, cloud-based platforms can now accept CT images, automatically segment the mandible, generate a family of implant designs using parametric rules, run finite element simulations on remote high-performance computing clusters, and return a validated design within hours instead of days.^{37,41} Integrated digital workflows may ultimately enable the rapid generation of implants optimized simultaneously for anatomical fit, manufacturability, and biomechanical performance. Automated design ecosystems integrating cloud computing, machine learning, and simulation-driven optimization have shown substantial promise for streamlining personalized implant production and enabling scalable digital healthcare manufacturing.

Such developments are expected to facilitate broader clinical adoption of patient-specific mandibular reconstruction systems and support the transition toward fully integrated digital reconstructive ecosystems. The convergence of AI, computational biomechanics, and AM may ultimately establish closed-loop digital reconstruction platforms capable of linking diagnosis, virtual planning, implant fabrication, and postoperative assessment within a unified clinical framework.

6. Conclusion

Additive manufacturing has fundamentally transformed the development of PSMRPs by enabling anatomically optimized implants with superior geometric complexity and biomechanical performance compared with conventional manually bent systems. Nevertheless, successful clinical translation of additively manufactured PSMRPs requires a comprehensive understanding of the interrelationship among manufacturing parameters, implant design, biomechanical behavior, and biological adaptation.

Current evidence demonstrates that LPBF-fabricated PSMRPs provide improved fixation accuracy, enhanced fatigue resistance, and more homogeneous stress distribution relative to conventional reconstruction plates. Structural optimization strategies, particularly topology optimization, offer substantial opportunities for reducing implant weight, mitigating stress concentration, and improving load transfer while maintaining biomechanical stability and functional durability.

Finite element analysis and experimental validation remain essential for evaluating implant safety, fixation stability, and long-term functional performance. Their integration has substantially improved predictive reliability and strengthened translational confidence in patient-specific mandibular reconstruction systems. Emerging developments involving AI-assisted design, mechanobiological simulation, and automated digital workflows are expected to accelerate the transition toward fully personalized mandibular reconstruction systems.

Ultimately, the convergence of AM, computational biomechanics, mechanobiological modeling, and digital surgical planning is reshaping mandibular reconstruction toward a future of integrated, patient-specific, and engineering-optimized craniomaxillofacial rehabilitation. Continued interdisciplinary collaboration will be essential for improving long-term implant reliability and advancing the clinical translation of next-generation patient-specific mandibular reconstruction systems.

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Conflict of interest

The authors declare they have no competing interests.

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