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Topology optimization of a mandibular reconstruction plate and biomechanical validation

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ABSTRACT

Objectives: Reconstruction plates, used to bridge segmental defects of the mandible after tumor resection or traumatic bone tissue loss, are subjected to repeated stresses of mastication. High stress concentrations in these plates can result in hardware failure. Topology optimization (TO) could reduce the peak stress by computing the most optimal material distribution in a patient-specific implant (PSI) used for mandibular reconstruction. The objective of this study was biomechanical validation of a TO-PSI.

Methods: A computer-aided design (CAD) model with a segmental defect was created based on the geometry of a polyurethane mandible model. A standard-PSI was designed to bridge the defect. A TO-PSI was then designed with a maximum stress equal to the ultimate tensile stress of Ti6Al4V (930 MPa) during a loading condition of 378 N. Finite element analysis (FEA) was used to analyze stresses in both PSI designs during loading. The standard-PSI and TO-PSI designs were produced in triplicate by selective laser melting of Ti6Al4V, fixated to polyurethane mandible models with segmental defects identical to the CAD model, and subsequently subjected to continuous compression with a speed of 1 mm/min on a universal testing machine, while recording the load. Peak loads before failure in the TO-PSI group within a 30% range of the predicted peak load (378 N) were considered a successful biomechanical validation.

Results: Fracture of the TO-PSI occurred at a median peak load of 334 N (range 304–336 N). These values are within the 30% range of the predicted peak load. Fracture of the mandible model in the standard-PSI group occurred at a median peak load of 1100 N (range 1010–1460 N). Failure locations during biomechanical testing of TO-PSI and standard-PSI samples corresponded to regions in the FEA where stresses exceeded the ultimate tensile strength of titanium and polyurethane, respectively.

Conclusion: This study demonstrates a successful preliminary biomechanical validation of TO in the design process for mandibular reconstruction plates. Further work is needed to refine the finite element model, which is necessary to ultimately design TO-PSIs for clinical use.

1. Introduction

Segmental defects of the mandible, for example after tumor resection, lead to severe disturbance in the form and function of the lower jaw. The loss of continuity is restored by bridging the defect with a titanium load-bearing reconstruction plate. Either a hand-bended reconstruction plate is used, or a more recently developed computer-designed patient-specific mandibular reconstruction plate (PSMP), otherwise called patient-specific implant (PSI) (Wilde et al., 2015). Ideally, the missing bone segment is reconstructed with an autologous

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bone flap (Rendenbach et al., 2018). In some cases, however, an osseous reconstruction is not possible because of comorbidities (Ferrari et al., 2013). In these cases, only a load-bearing reconstruction plate is used to bridge the defect between the two mandibular stumps.

Since these plates are subjected to repeated stresses of mastication, hardware-related complications occur frequently. Plate fracture rates up to 8.3% have been reported (Shibahara et al., 2002; Katakura et al., 2004). Whereas screw fractures have occurred but are rarely reported (Ueyama et al., 1996; Siegmund et al., 2000), screw loosening does take place when the cortical bone directly surrounding the screws is overloaded by the stresses, leading to bone resorption (Knoll et al., 2006). Next, radiation therapy was found to be associated with an increased risk of plate exposure (Fanzio et al., 2015). These hardware-related complications often require the removal or replacement of the load-bearing reconstruction plate in an additional operation, thereby significantly increasing the patient's morbidity.

Finite element analysis (FEA) and scanning electron microscopy have shown that plate fractures are caused by high stress concentration regions in the reconstruction plates (Katakura et al., 2004; Al-Ali et al., 2017). In addition, repetitive pre- or intraoperative bending of load-bearing reconstruction plates to fit the patient's mandible is thought to cause weak spots. These spots are prone to fracture when the plate is loaded during normal mastication (Martola et al., 2007). Several studies have focused on improving the design of the load-bearing reconstruction plates (Knoll et al., 2006; Huo et al., 2015; Jo et al., 2018; Gateno et al., 2013; Rendenbach et al., 2017, 2019; Zimmermann et al., 2017; Bujtar et al., 2014; Wong et al., 2010; Qassemyar et al., 2017; Kumar et al., 2015; Wu et al., 2017). An interesting approach is the use of PSIs instead of conventional hand-bended titanium plates. PSIs are based on the patient's anatomy using medical imaging and computer-aided design (CAD) software, and are subsequently manufactured from titanium with additive manufacturing (AM) techniques (Wilde et al., 2015; Cornelius et al., 2015). PSIs have shown to provide predictable surgery, which eventually aids dental rehabilitation of the patient (Schepers et al., 2016). Furthermore, it is argued that PSIs are less prone to fracture, since they are not subjected to repetitive pre- and intraoperative bending to fit the plate to the patient's mandible. However, the current PSIs are not designed based on complete biomechanical analysis and their dimensions are frequently similar to conventional hand-bended reconstruction plates. As a result, regions with high stress concentrations could still be present, leading to the aforementioned complications (Gutwald et al., 2017).

Recently, researchers started using the computational method called topology optimization (TO) for the conceptual design of implants in multiple clinical fields, such as orthopedics, cardiothoracic surgery, otolaryngology, and in cranio-maxillofacial (CMF) surgery to develop new mandibular reconstruction plates (Seebach et al., 2018; Lemon, 2016; Sigmund, 2000; Rahimizadeh et al., 2018; Khalighi et al., 2019; Milazzo et al., 2020). TO is a mathematical method that makes use of finite element (FE) models to optimize the material distribution within a set geometric design space and given a certain loading configuration as input (Sigmund, 2000; Deaton and Grandhi, 2014; Zhu et al., 2016). The method is different compared to shape optimization as it does not limit geometry within the design space. The method has been successfully applied in several engineering domains, for example to reduce the weight of truss structures of bridges and buildings. Recent studies show promising TO designs of bone plates for mandibular fracture repair with a significant decrease in peak stresses when compared to standard osteosynthesis plates (Lovald et al., 2009, 2010).

We hypothesize that an individual reconstruction plate design, tailored to the needs of a specific patient using TO, can reduce the incidence of hardware-related complications. To date, none of the TO designs for mandibular reconstruction have been validated using biomechanical tests. Thus, the objective of this study is biomechanical validation of TO mandibular reconstruction plates.

2. Materials and methods

First, we generated a simulated mandible model with a virtual segmental defect. Next, we designed a standard PSI (standard-PSI) and a topology optimized PSI (TO-PSI) for this mandibular defect model. Finally, both designs were manufactured in triplicate, fixated to polyurethane mandibular defect models and subjected to a continuous compression test.

2.1. Mandible model with segmental defect

A CAD mandible model was created based on the geometry of a commercially available mandible replica (Mandible 8950, Synbone, Malans, Switzerland) (Fig. 1A). A replica was chosen instead of a human mandible to enable repeatable biomechanical testing. The mandible replica was made of polyurethane with an outer part representing the cortex and an inner part representing cancellous bone (Rendenbach et al., 2017, 2019; Bredbenner and Haug, 2000; Schupp et al., 2007) (Fig. 1B).

A photogrammetric measurement scan (GEOSCAN, Leiden, the Netherlands) and a computed tomography (CT) scan were made of the mandible replica. The replica was positioned on a turntable during the photogrammetic measurement scan, which was 15 cm away from two 5 megapixel cameras resulting in a scan precision of 8–10 μm (Spectrum Premium, Rangevision, Moscow, Russia). The CT scan was made in-house with a slice thickness of 0.5 mm, 120 kV/23 mAs and a field of view of 512 × 512 pixels.

The outer shape of the CAD model was established using the photogrammetric scan data because of its high resolution, resulting in a solid volume. The CT scan was used to determine the internal cancellous bone volume. Both the outer shape and the cancellous bone volume were combined using Creo 3.0 (PTC, Boston, USA), resulting in a representation of the cortical thickness. The volume of the complete mandible in the CAD model was 58.3 cm^3. A right-sided mandibular defect type that is commonly encountered in clinical practice was created, corresponding to a Class II defect as described by Brown et al., including both a horizontal and a vertical corner and crossing the midline of the mandible (Fig. 1A) (Brown et al., 2016). The volume of the segment of the mandible defined as the defect in the CAD model was 25.5 cm^3.

Standard values for the mechanical properties of polyurethane (i.e. Young's modulus, Poisson's ratio and ultimate tensile strength) were assigned to the cortical part of the model (Table 1) (Materialise, 2020). The inner cancellous part was left as a cavity, since its relatively low stiffness would have a negligible effect on the stress distribution in the reconstruction plate (Huo et al., 2015). In the FE model load and support conditions were set as follows. Movement of the most anterior remaining teeth was restricted to the horizontal plane. The right condyle was only allowed to move laterally or medially in the horizontal plane, while the left condyle was restrained from movement in all directions (Fig. 2A). Load was applied on both mandibular angles and the mandible was positioned with its inferior border oriented at 8° to the horizontal plane, so the resulting vector of the mastication muscles could be simulated (Seebach et al., 2018).

2.2. Reconstruction plate design

2.2.1. Standard-PSI and modelling

A standard-PSI was designed in an online planning session with a commercial manufacturer (Xilloc, Sittard-Geleen, the Netherlands). AO Foundation principles on osteosynthesis were followed for the design (Cienfuegos et al., 2008). The design was exported into a STEP-file (STandard for the Exchange of Product model data) for transfer to Creo 3.0 and was then added to the virtual mandible model (Fig. 3A). The mechanical properties of titanium grade 5 were assigned to the standard-PSI plate and screws (Table 1). Each screw shaft was modelled as a solid cylinder with a diameter of 2.0 mm, extending into the lingual

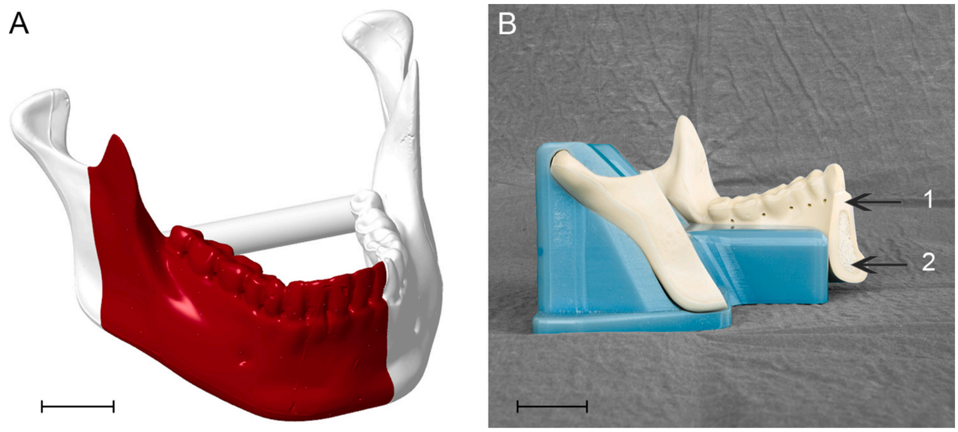

Fig. 1. Mandible model with segmental defect. A) CAD model of a mandible replica with segment planned for resection shown in red. B) Polyurethane mandible model representing cortical bone (arrow 1) and cancellous bone (arrow 2), with segmental defect, in positioning jig. Scale bars: 20 mm.

Table 1
Mechanical properties assigned to the FE model.

	Young's modulus (GPa)	Poisson's ratio	Ultimate tensile strength (MPa)
Polyurethane	0.5	0.3	69
Titanium grade 5 (Ti6Al4V)	114	0.3	930

cortical section. All screws were inserted and perfectly bonded along their surface to the cortical sections using Boolean operations. The screw-plate interface was modelled as bonded as well, and the plate-mandible interface was set as free without any contact elements.

2.2.2. Topology Optimized-PSI

Using the standard-PSI design as the predefined geometric design domain, a TO-PSI was developed using ProTOp 5.0 (CAESS, Maribor, Slovenia).

For the experimental biomechanical validation in our current study, we pursued to design an optimized reconstruction plate which was expected to fracture well before failure of the polyurethane mandible model would occur. Therefore, the design goal was set to a maximum stress equal to the ultimate tensile strength of titanium (Table 1) and the design loading condition was set at a relatively low load of 378 N. This design loading condition was derived from a previous experimental test in which a polyurethane mandible model was cyclically loaded to mimic mastication. The polyurethane mandible model fractured at a load of 540 N and by application of a safety factor of 0.7 this generated our design loading condition of 378 N. Please notice that the loading value is not the value that would be selected for clinical application, as then the human cortical bone would dictate the optimization criterion.

The geometric regions that were excluded from the design domain included a circular area around each screw hole and the part of the standard-PSI plate adjacent to the mandibular segments needed for fitting the plate during surgery (Fig. 3B). The inferior and superior geometric borders of the standard-PSI design were set as a fixed region, to create a shell-like shape (Fig. 3C). The remaining part of the design space was configured as a lattice infill design with rods as a base shape

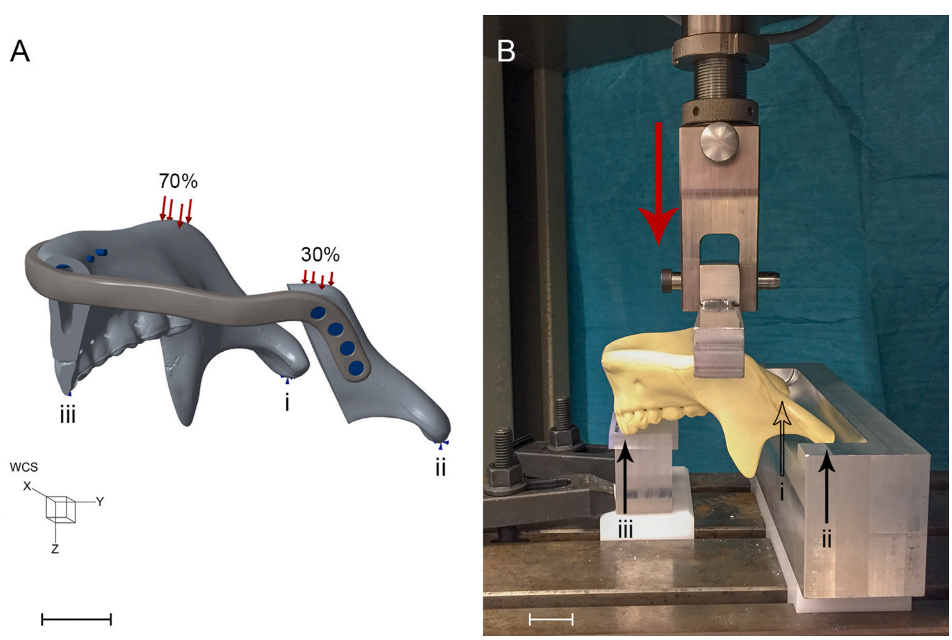

Fig. 2. FEA model and biomechanical testing set-up. A) Mandible replica with arrows representing the 70%–30% distribution of the load on the mandibular angles. i) Movement of left condyle restricted in all directions, ii) Movement of right condyle only allowed laterally or medially in horizontal plane, iii) Movement of anterior teeth restricted to horizontal plane. B) Polyurethane mandible model positioned in testing jig following the FEA model setup. Scale bars: 20 mm.

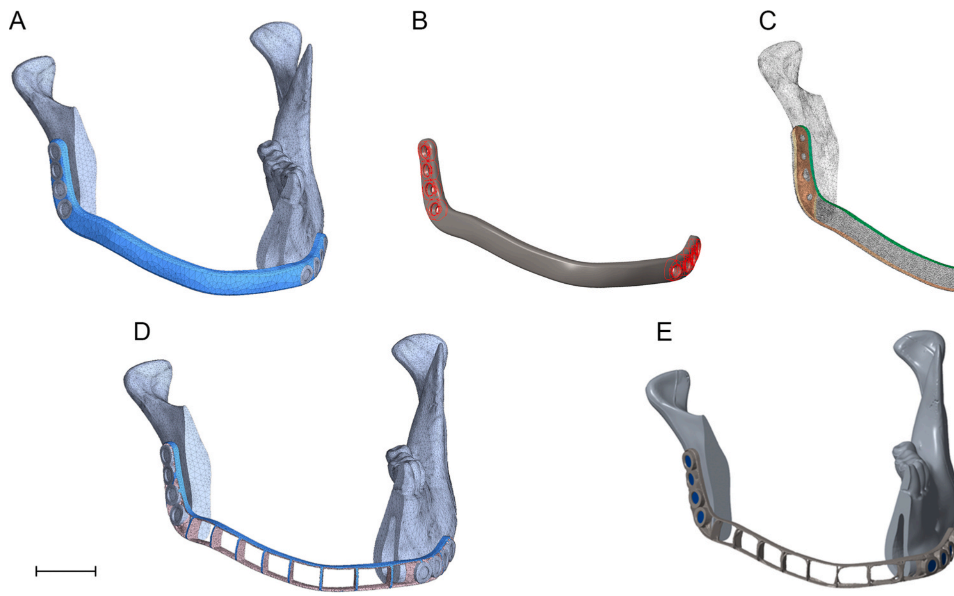

Fig. 3. Design of TO-PSI. A) Standard-PSI as predefined geometric design domain. B) Circular area around screw holes excluded from design domain. C) Inferior and superior borders as fixed regions. D) Rods as lattice infill design. E) Final TO-PSI design. Scale bar: 20 mm.

(Fig. 3D), since this shape would make it easier to compare the expected fracture location in the FEA with the fracture location of the experimental biomechanical testing. A minimum rod size of 0.5 mm was set to prevent manufacturing problems. The final TO-PSI design after 32 iterations is shown in Fig. 3E.

2.3. Finite element analyses

FEA of the mandible model with the standard-PSI and TO-PSI designs were performed in Creo 3.0 to determine the load and location of stresses exceeding the ultimate tensile strength of titanium or polyurethane with the set conditions.

2.4. Biomechanical testing

2.4.1. Polyurethane models

The standard-PSI and TO-PSI designs were produced in triplicate by AM using selective laser melting of titanium alloy (titanium grade 5; Ti6Al4V) (Fig. 4). Identical segmental defects equal to the CAD mandibular defect model were created in six polyurethane mandible models (Fig. 1B), using a circular saw and saw template. Subsequently, the standard-PSIs and TO-PSIs were fixated to the mandible segments using non-locking bicortical screws of 10–18 mm length and a diameter of 2.7 mm (25-886-10, −16, −18, KLS Martin, Tuttlingen, Germany) using a positioning jig to ensure identical positioning of the mandibular segments. The screw lengths related to their positions were identical in all six mandible models. Sawing of the mandible models and fixation of the PSIs was performed by DK.

2.4.2. Test set-up and protocol

We aimed for biomechanical validation of our TO design method and subsequent FEA. For this, we chose the experimental protocol, as described below, to be leading. That is: loading till failure was applied, as has been performed in similar studies (Rendenbach et al., 2017, 2019; Gutwald et al., 2017). Therefore, this same loading protocol and boundary conditions were applied to perform the FEA. This approach made it possible to verify the outcome of the FEA with the experimental results, and thus a first validation of our computational design method. A universal testing machine (1440 Zmart Pro, Zwick/Roell, Ulm, Germany) was used for biomechanical testing. A custom-made jig was designed to position the polyurethane mandible models in the machine (Fig. 2B). The models with the fixated PSIs were loaded in a reverse position for practical reasons. Following Schupp et al., 70% of the load was allocated to the intact left-hand side of the mandible while the remaining 30% was distributed to the resected right-hand side using a see-saw device (Schupp et al., 2007). The mandible models with the fixated PSIs were subjected to continuous compression testing with a speed of 1 mm/min. Pre-load was set at 20 N and the maximum load was set at 1600 N. As a safety precaution, the compression test would stop automatically when a deformation of 20 mm was exceeded or when the load would suddenly decrease to less than 50% of what was applied by the testing machine.

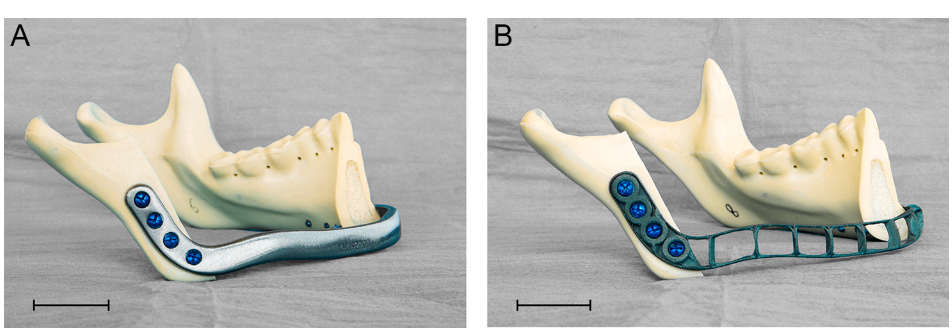

Fig. 4. Polyurethane mandibular defect models with fixated PSIs. A) Standard-PSI. B) TO-PSI. Scale bars: 20 mm.

2.4.3. Data collection and outcomes

Time (s), load (N) and vertical displacement (mm) were recorded using the universal testing machine. A GoPro Hero 6 (GoPro, San Mateo, USA) was used for continuous image analysis.

The primary outcome variable was peak load (N) before failure of the reconstruction plate, screws or mandible. Failure was defined as a vertical displacement of 20 mm or more, or a visible fracture of the reconstruction plate, screws or mandible as recorded by the GoPro.

The raw load-displacement data were graphed, and exploratory data analysis was performed using Microsoft Excel, Version 14.0.7232 (Microsoft, Redmond, USA). Peak loads and failure locations of the experiments were compared to the predicted peak loads and failure locations in the FEA. Peak loads before failure in the TO-PSI group within a 30% range of the predicted peak load were considered a successful biomechanical validation.

3. Results

FEA of the TO-PSI design confirmed the predicted peak load of 378 N. Multiple regions showed peak stresses that exceeded the ultimate tensile strength of titanium (Fig. 5A).

During biomechanical testing, failure in the TO-PSI group occurred at a median peak load of 334 N (range 304–336 N), which was within a 30% range of the predicted peak load of 378 N. In all three TO-PSI samples, the mechanism of failure was a fracture in the right paramedian region of the PSI (Fig. 5B), which corresponded to a peak stress location in the FEA (Fig. 5A). Failure in the standard-PSI group occurred at a median peak load of 1100 N (range 1010–1460 N). The mechanism of failure for all samples in this group was fracture of the mandibular ramus on the right-hand side of the polyurethane mandible models, involving one or more screw holes (Fig. 5D), which corresponded to regions in the FEA where stresses exceeded the ultimate tensile strength of polyurethane (Fig. 5C).

Image analysis of the GoPro data confirmed failure in both groups. None of the specimens reached a vertical displacement of 20 mm before any other mechanism of failure occurred. The steeper slopes of the force-displacement curves in the standard-PSI group indicated a higher stiffness of these PSIs compared to the TO-PSI group, as expected (Fig. 6).

4. Discussion

This study set out to validate the design process using TO for mandibular reconstruction plates in a representative biomechanical testing set-up. We designed a TO-PSI that reached the ultimate tensile strength of titanium at a load of 378 N. Our results of biomechanical testing show a successful preliminary validation with peak loads before failure occurring within a 30% range of the predicted peak load, and fracture locations matching the regions with peak stresses in the FEA. Likewise, the fracture location of the mandible models in the standard-PSI group also matched the peak stress locations in the FEA. These results suggest that our FE modelling and chosen conditions sufficiently match reality to apply them for the design of mandibular reconstruction plates, where the FEA allows reliable prediction of the peak loads and locations. A distinct example is shown in this study, where the absence of fractures at supraphysiological peak loads for the standard-PSI plates suggest that they can benefit from removal of excess material to optimize the shape of the plate and bring biomechanical behavior of the plate in line with that of the bone, and as a result reduce peak stress at the fixation locations.

FEA is a convenient and widely accepted modelling tool (Trivedi, 2014). In order to compare TO-PSI designs, FEA could be used as a first screening to highlight the most promising designs. It allows changing of many parameters such as mechanical properties in order to quickly analyze the effects on the design and peak loading behavior. Furthermore, it is easily repeatable and lacks ethical restraints. Despite these advantages, we should bear in mind that FEA is only a model of reality.

TO has only recently been introduced to develop new mandibular reconstruction plates. The design algorithm includes several steps. First, the design and non-design domains have to be defined by the engineer on the conventional design structure. The non-design space in a mandibular reconstruction plate will be the areas around the screw holes, for example. Similarly, the load and support conditions that will

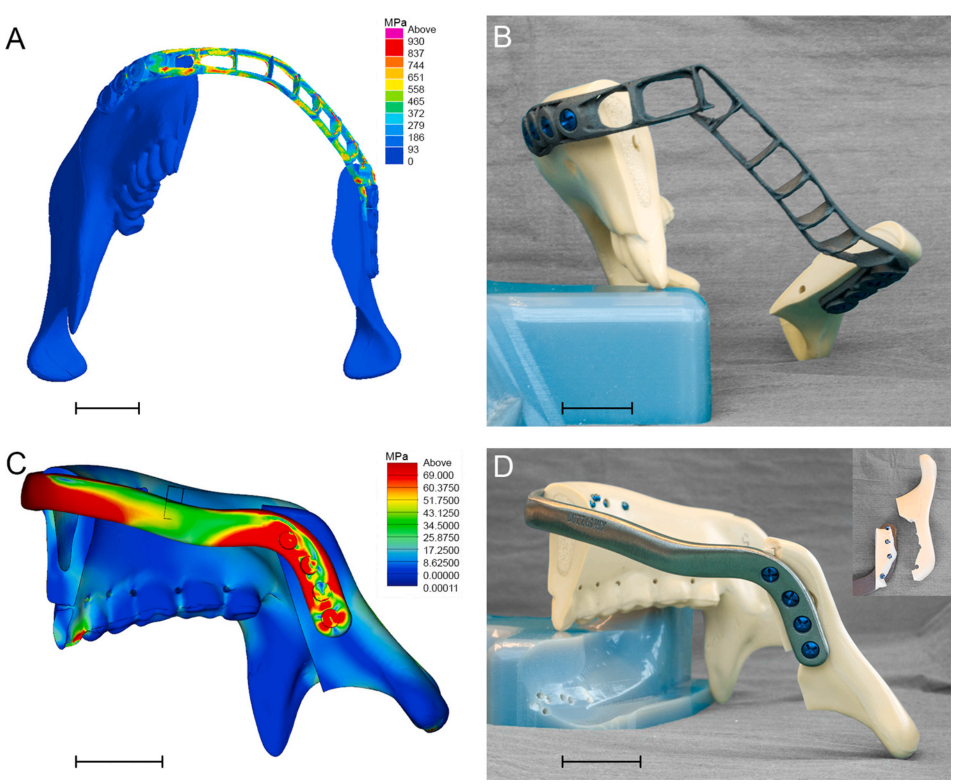

Fig. 5. FEA and biomechanical test results. A) Stress distribution in TO-PSI at a load of 378 N. Red regions represent stresses above the ultimate tensile strength of titanium. B) Fracture of a TO-PSI located in the right paramedian region. C) Stress distribution in standard-PSI at a load of 1000 N. Red regions represent stresses above the ultimate tensile strength of polyurethane. D) Fracture of the polyurethane mandible model in the standard-PSI group, involving the screw holes. Insert showing a medial view of the fracture. Scale bars: 20 mm.

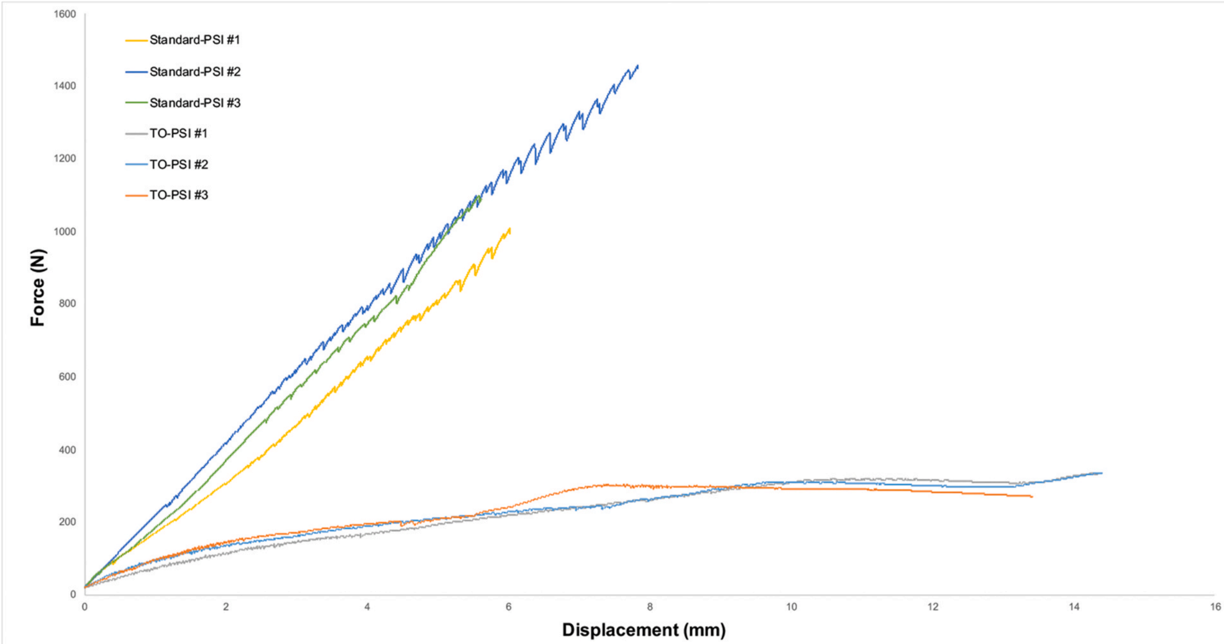

Fig. 6. Force-displacement curves of standard-PSI samples and TO-PSI samples during biomechanical testing. Each line represents an individual sample tested.

influence the design of the structure have to be defined, together with the material properties of each structure. In mandibular reconstruction, the loads will be the bite forces exerted by the masticatory system. With these data, the algorithm calculates the response of the conventional design structure using FEA. This analysis is the basis for improving the design. The structure is subdivided into smaller elements to eventually combine their equations into a response of the total structure. Then, the engineer has to define the optimization goals. A goal, for example, could be to decrease the maximum stress in the structure below the known breaking strength of the material, so fracture will less likely occur. Similarly, decreasing maximum stresses around the screws could prevent screw loosening. Finally, the optimization algorithm computes the most optimal material distribution to meet these goals.

Our FE model represents a three-point loading model with restrained movement of both condyles and left-hand side canine region during loading of the mandibular angles. The test set-up was then designed to closely match the FE model. The three-point loading model was regarded the most suitable for the present study since it was the most feasible to reproduce in a test set-up. Other studies evaluating mandibular reconstruction plates have also used a similar three-point loading set-up. (Rendenbach et al., 2019; Gutwald et al., 2017). Rendenbach et al. found that their titanium milled standard-PSIs did not fracture during biomechanical testing up to a maximum load of 1000 N (Rendenbach et al., 2019). In contrast, Gutwald et al. observed plate fracture of laser sintered standard-PSIs at a load of 700 N (Gutwald et al., 2017).

We chose to use polyurethane models in this study. In these polyurethane models all cortical regions have the same mechanical properties, while the mechanical properties of the human mandibular cortex vary between regions (Schwartz-Dabney and Dechow, 2003). The mechanical properties of these polyurethane models have been shown to be roughly comparable human bone (Bredbenner and Haug, 2000). A number of previous studies investigating the biomechanics of mandibular reconstruction plates have also used polyurethane mandible replicas (Rendenbach et al., 2017, 2019; Schupp et al., 2007de Medeiros et al., 2016). Since the goal of our study was the biomechanical validation of a promising technique, we have chosen to use readily available identical polyurethane models, since these increase the internal validity and reproducibility of the experiment. In addition, ethical, practical and financial obstacles, known in study designs using human specimens, were avoided (Bredbenner and Haug, 2000). The purpose of the TO is to lower the stiffness of the PSI as compared to the standard-PSI and to match the stiffness of bone. When the PSI stiffness is in the same range of bone, this contributes to minimization of stress shielding. Since we did not use actual bone in the biomechanical validation, but a polyurethane mandible model, we matched the TO design to the strength and stiffness of polyurethane. Further research will be carried out to explore a method to calculate mechanical properties of an individual human mandible using standard medical imaging techniques (Wagner et al., 2011). These patient-specific mechanical properties of mandibular bone are necessary to ultimately design TO-PSIs for clinical use.

One limitation of our study is the small sample size of three samples per group. However, the intra-group variation was kept as low as possible by using standardized polyurethane mandible models, a saw template to create identical segmental resections, a positioning jig during fixation of the PSIs, the same CMF surgeon performing all procedures, and identical positions in the testing machine. As a result, the peak loads before failure, represented by the force-displacement curves, show little variation within the groups. Therefore, we believe our study is an accurate biomechanical validation of the TO-PSI despite its small sample size.

The masticatory system is highly complex. Movement of the lower jaw in physiological conditions is generated by 16 jaw muscle groups and knows many degrees of freedom (Peck, 2016). Because of the complexity of this biomechanical system, simplification is necessary for experiments. A simplified resulting vector of bite force has been reported in the literature and was applied in our experiments (Seebach et al., 2018). It has also been reported before that after resection of a segment of the mandible, the contribution of force used for biting is not equally distributed on both sides of the mandible (Marunick et al., 1992; Curtis et al., 1999). Therefore, we assigned 70% of the contribution to the intact side, and 30% to the defect side (Schupp et al., 2007). Ultimately, loading for validation in our study is different from functional mastication loading. It may be advocated that a cyclic loading protocol is considerably more similar to the normal loading of the masticatory system than the application of a static load (Zimmermann et al., 2017). Several other studies evaluating mandibular reconstruction plates have indeed used a cyclic loading protocol (; Zimmermann et al., 2017; Rendenbach et al., 2017; Schupp et al., 2007;). Nevertheless, a static loading test was more suitable for the validation of our FEA, since current commercially available FEA software, including Creo 3.0, mainly focusses on the analysis of static loading (Swierstra, 2017).

In biomechanical modelling, uncertainties are present. We minimized these uncertainties by using a relatively simple testing protocol, well-defined loading and fixed conditions, a well-defined bone defect size, and the use of a polyurethane mandible model which allowed us to design an accurate computational model with the use of CT-segmentation. Since we had a low number of experiments, further statistical analyses were not possible. However, an uncertainty analysis could be used in future experiments with greater sample sizes (Hamdia et al., 2019).

TO allows for a wide variety of PSI designs, while securing the strength of the PSI including a safety margin, since one can vary both the optimization goals and geometric design domains. This could enhance clinical outcomes. Known complications like plate or screw fracture can be avoided by reducing maximum stresses in the material. Plate exposure after radiotherapy, which could partly be related to plate thickness, might be avoided by designing flush PSIs using TO. (Fanzio et al., 2015). In addition, a macroporous PSI could be designed to promote ingrowth of soft tissues in patients where osseous reconstruction is impossible because of comorbidities (Qassemyar et al., 2017). Finally, the use of other biomaterials for mandibular reconstruction could be explored using TO designs (Zhang et al., 2019).

5. Conclusion

This study demonstrates successful biomechanical validation of a TO design process for mandibular reconstruction plates in an experimental set-up. Our results indicate that TO with FEA can be used as a versatile CAD process for mandibular reconstruction plate design. Further work is needed to explore methods to calculate individual mechanical properties of human mandibular bone, and to closely match the loading conditions of actual human mastication. This will result in a refined FE model, which is necessary to ultimately design TO-PSIs for clinical use.

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CRediT authorship contribution statement

David C. Koper: Conceptualization, Investigation, Formal analysis, Writing - original draft. **Carine A.W. Leung:** Investigation, Formal analysis, Data curation, Writing - original draft. **Lars C.P. Smeets:** Investigation. **Paul F.J. Laeven:** Resources. **Gabriëlle J.M. Tuijthof:** Methodology, Writing - review & editing. **Peter A.W.H. Kessler:** Conceptualization, Supervision, Funding acquisition.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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List of abbreviations

AM: Additive Manufacturing
CAD: Computer-Aided Design
CMF: Cranio-Maxillofacial
CT: Computed Tomography
FE: Finite Element
FEA: Finite Element Analysis
PSI: Patient-Specific Implant
PSMP: Patient-Specific Mandibular Reconstruction Plate
STEP: STandard for the Exchange of Product model data
TO: Topology Optimization

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