Fracture Prediction for a Customized Mandibular Reconstruction Plate with Finite Element Method

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Abstract. The use of customized reconstruction plate is an effective method for reconstruction of mandibular continuity defects. Plate fracture is one of the most common postoperative complications. The aim of this study was to investigate the biomechanical behavior of the customized reconstruction plate by finite element method. The geometry model was created from computed tomography (CT) data of a patient. The muscle forces for the defected mandible under two common static biting tasks were estimated by a numerical optimization strategy with the objective function of minimization of overall muscle force. The simulation results revealed that changing bite from molar region to incisor region increased the maximum stress in the plate. The position of stress concentration, the upper-inner edge of the plate near ramus-end, was in agreement with that of fracture, which indicated that stress concentration regions were critical regions for fracture failure.

Keywords: Finite element analysis · Muscle forces · Reconstruction plate · Fracture

1 Introduction

Mandible defects without timely reconstruction can lead to disturbed mastication, impairment of speech, and facial deformity, which seriously affect patients' quality of life. The objective of mandibular reconstruction is to restore both the shape and the function of the mandible.

The use of titanium reconstruction plate with or without bone grafting has become a popular choice for repairing mandibular defects. Standard titanium reconstruction plates are commonly used in surgery. Currently, with the development of computer-aided design techniques and additive manufacturing techniques, customized titanium reconstruction plates with specific geometrical shapes can be fabricated and are gradually used in oral maxillofacial surgery. Compared with standard reconstruction plates, customized plates offer many benefits, including better bone surface adaption and superior facial recovery. Nevertheless, there are some serious long-term complications after plate implantation surgery, such as plate exposure, plate fracture and screws loosening.

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According to the previous literature, incidence of plate fracture ranges from 2.9% to 10.7% [1]. Plate fracture mostly occurs in less than 6–9 months after surgery. It is more frequent in patients with a bone resection including the mandibular angle, most commonly occurring near the anterior region of the mandibular angle [2]. By material analysis, it was observed that the origin of cracks was stress concentration regions in the plate and metal fatigue was caused by frequently masticatory function. Thus, it becomes important to validate the biomechanical behavior of the reconstruction plate.

The objective of this study was to estimate reasonable muscle forces of the defected mandible with an optimization algorithm and to investigate the biomechanical behavior of the reconstruction plate under masticatory loads with finite element method. The position of the plate fracture and that of the stress concentration were also compared.

2 Materials and Methods

A 45-year-old female patient with ameloblastoma was chosen for this study. She got extensive tumor resection and immediate mandibular reconstruction surgery with a customized titanium plate. The customized titanium reconstruction plate extending from the chin to the mandibular angle was anchored to the residual mandibular bone with seventeen 2.7-mm-diameter bicortical osteosynthesis screws. Unfortunately, plate fracture occurred in several months after surgery.

2.1 Generation of the Geometric Model

The anatomical geometry of the mandible with defects was reconstructed from postoperative CT data of the patient by using MIMICS (Version 10.01, Materialise, Inc.). By employing threshold segmentation method, regions of bone and titanium plate were separated and reconstructed separately. Screws were modeled as 2.7-mm-diameter cylindrical pins in Geomagic Design X (Version 2016. 0. 1, 3D systems, Inc.). After position aligning, boundary smoothing and volume forming, the geometric model of the assembly of plate/mandible/screw system was obtained in Geomagic Studio (Version 12.0, Geomagic, Inc.).

2.2 Computation of Muscle Forces

The two most common chewing tasks, i.e. incisal clenching (INC) and left molar clenching (LMOL) were simulated in this study. Due to the segmental mandibular resection, a reduced biting force of 300 N in the vertical direction was chosen. The extensive mandiblular resection (including the mandibular angle) meant that right masseter muscles were destroyed. To determine reasonable muscle and joint forces of the defected mandible, an optimization strategy with the objective function of minimization of overall muscle force was carried out [3]. Seven pairs of major masticatory muscles, namely the superficial masseter, the deep masseter, the anterior temporalis, the middle temporalis, the posterior temporalis, the lateral pterygoid and the medial pterygoid, with the absence of right masseter muscles were included in this study. The joint load was set as a single component with a constrained direction. The direction of joint force was considered to be normal to the joint surface at the articular contact, which was measured in the geometric model.

The resultant bite force, muscle and joint forces must fulfill the following six equilibrium conditions:

$$\sum_{i=1}^{12} \mathbf{M}_i + \mathbf{J}_r + \mathbf{J}_l + \mathbf{B} = \mathbf{0} \tag{1}$$

$$\sum_{i=1}^{12} (\mathbf{r}_i \times \mathbf{M}_i) + (\mathbf{r}_{jr} \times \mathbf{J}_r) + (\mathbf{r}_{jl} \times \mathbf{J}_1) + (\mathbf{r}_b \times \mathbf{B}) = \mathbf{0} \tag{2}$$

where $\mathbf{M}_i$, $\mathbf{J}_r$, $\mathbf{J}_l$ and $\mathbf{B}$ represent muscle force, right joint force, left joint force and vertical bite force, respectively. The magnitude of vertical bite force was set to be 300 N in incisor region and left molar region successively, for simulating incisor clenching and left molar clenching.

The magnitude of muscle and joint forces must fulfill the following constraint conditions:

$$0 \le |\mathbf{J}_r| \text{ and } 0 \le |\mathbf{J}_l| \tag{3}$$

$$M_\mathrm{i} = |\mathbf{M}_i| \text{ and } 0 \le M_i \le m_\mathrm{i} \tag{4}$$

where m_i represents maximum tensile force of each muscle.

During optimum processing, muscle forces were determined when the following objective function achieved. Afterward, these calculated muscle forces were applied in finite element analysis.

$$f = \min[\sum_{i=1}^{12} M_i] \tag{5}$$

2.3 Finite Element Analysis

The geometric model of the assembly of plate/mandible/screw system was imported into ANSYS (Version 14.0, ANSYS, Inc.) for mechanical analysis. The finite element model was shown in Fig. 1. The model was meshed with linear tetrahedrons elements. The mesh consists of 799,327 elements and 151,045 nodes, which was dense enough to ensure calculation accuracy. A bond-type connection was applied to the interface of locking screws and the reconstruction plate. Material properties involved in the model were all considered isotropic, homogeneous and linear elastic (Table 1).

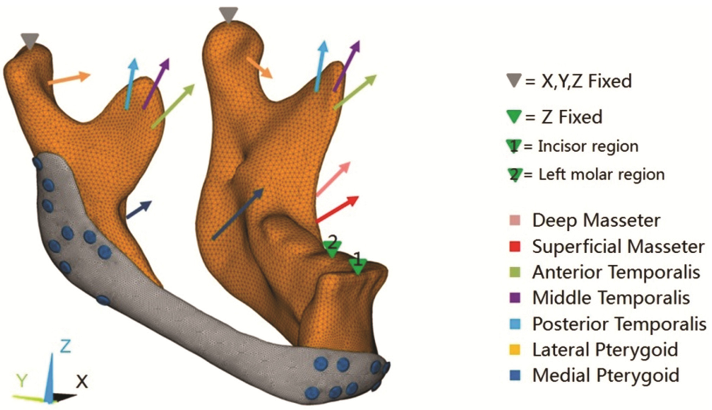

Fig. 1. Finite element model of the assembly of the mandible and the reconstruction plate. Boundary conditions and loads are also briefly illustrated.

Table 1. Material properties of different parts in the finite element model.

Type of material	Young's modulus [MPa]	Poisson's ratio
Mandible (cortical bone)	8700	0.28
Ti6Al4 V (plate and screw)	105,000	0.30

Incisor clenching and left molar clenching were simulated by applying corresponding muscle forces to the mandible. The top surfaces of two condyles were fully restrained to prevent the rigid-body displacement of the mandible. Displacement in vertical direction of corresponding occlusal contacts was constrained too. Values of muscle forces were obtained from calculated results.

3 Results

Calculated muscle forces and joint forces are shown in Table 2. Joint force can only be transmitted by compression and its direction was normal to the joint surface at the articular contact. On the basis of the condyle morphology, the measured joint force vector pointed back and downward in the saggital plane with an angle of 26° away from vertical. Under both chewing tasks, the superficial masseter, the anterior temporalis and the medial pterygoid were the most heavily loaded muscles, and the superficial masseter was proportionally larger than the medial pterygoid. Deep masseter and posterior temporalis were not active during any biting task. It seems that the superficial masseter was always activated in preference to the deep masseter. During incisor clenching, due to the absence of right masseter muscles, the right and the left muscle groups were not recruited equally. During left molar clenching, muscles on the balancing side except the anterior temporalis were totally suppressed. It was observed that the joint force at the left side was larger than that at the right side under both loads. In addition, changing bite from incisor region to left molar region largely reduced joint loads.

Table 2. Calculated muscle force, joint forces and corresponding directions.

	Max. force (N)	Calculated muscle force (N)				Direction of force		
		INC		LMOL		Cos-x	Cos-y	Cos-z
		Right	Left	Right	Left			
Superficial Masseter	190.4	–	190.4	–	129.1	−0.21	−0.42	0.88
Deep masseter	81.6	–	0	–	0	−0.55	0.36	0.76
Medial pterygoid	174.8	123.4	174.8	0	93.9	0.49	−0.37	0.79
Lateral pterygoid	66.9	0	25.2	0	0	0.63	−0.76	−0.17
Anterior temporalis	158.0	158.0	158.0	89.1	158.0	−0.15	−0.04	0.99
Middle temporalis	95.6	9.5	16.1	0	39.0	−0.22	0.50	0.84
Posterior temporalis	75.6	0	0	0	0	−0.21	0.86	0.47
Joint force	–	141.8	340.3	42.8	141.0	0	0.44	−0.90

The displacement distribution in the reconstructed mandible under incisor clenching and left molar clenching was analyzed (Fig. 2). For better display, the deformation was magnified by three times in Fig. 2. Unsymmetrical displacement distribution was observed on the mandible under both biting tasks. Specifically, the displacement on the defected side of the mandible was larger than that on the opposite side. The maximum displacement in the plate was 0.83 mm under incisor clenching, decreasing to 0.25 mm under left molar clenching. Under incisor clenching, the plate slightly moved towards the left side. The largest displacement occurred close to the resection site on the right ramus end. When the biting position moved to left molar, displacements on both sides of the mandible decreased and the largest displacement was observed in the overhanging part of the plate.

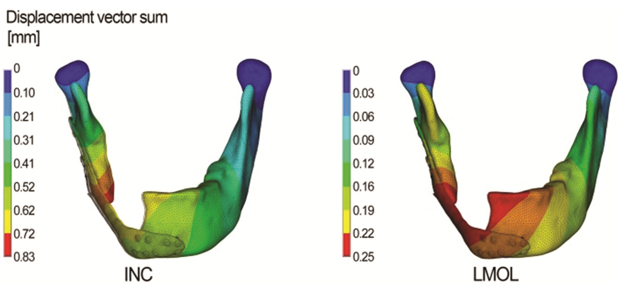

Fig. 2. Total displacement [mm] in the mandible and the plate during INC and LMOL. The undeformed model is meshed with black lines and the deformed model displays in color.

Figure 3 shows the von Mises stress distribution in the reconstruction plate during incisor clenching and left molar clenching. Under both clenching tasks, the patterns of stress distribution in the plate were quite similar: the overhanging part of the plate was more critically loaded than two fixed ends and the maximum von Mises stress in the plate was both observed on the inner-upper edge near the ramus end-incisor clenching: 260.55 MPa and left molar clenching: 92.91 MPa. Another high stress concentration region in the plate was also observed on the inner-lower edge near the chin end with a stress of approximately 233 MPa. The screw holes near the resection sites exhibited higher stress. Figure 4 shows the fractured plate reconstructed geometrically from the CT data after plate failure, in which the fracture of the plate is highlighted by a red circle. The position of the plate failure and that of the largest von Mises stress in the plate calculated by FEM agreed well.

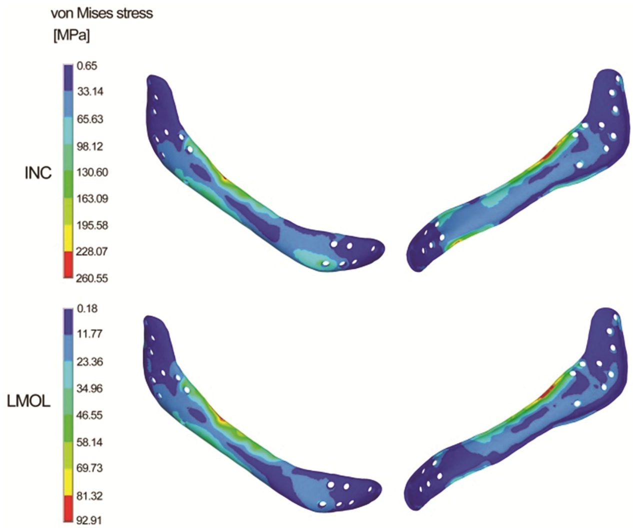

Fig. 3. von Mises stress [MPa] in the reconstruction plate during INC and LMOL.

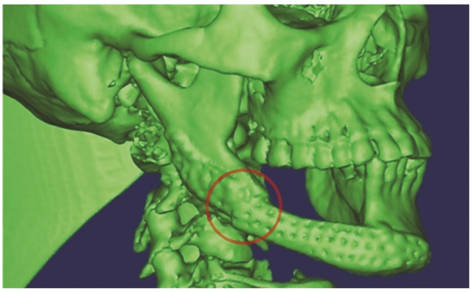

Fig. 4. The fractured plate reconstructed geometrically from the CT data after plate failure. (Color figure online)

The maximum and minimum principal strain distributions in the mandible were shown in Fig. 5. In general, the screw holes near the resection sites on the chin and ramus ends exhibited higher strain. The upper ramus-end hole was considered the most critical area as it always demonstrated the highest tension and compressive strain under both loading tasks.

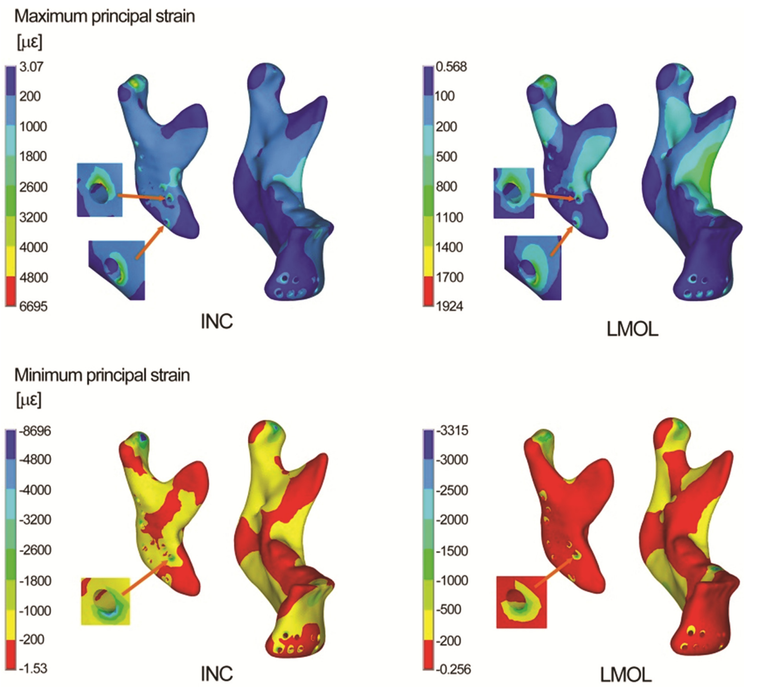

Fig. 5. Maximum and minimum principal strain [με] in the mandible during INC and LMOL.

4 Discussion

The female patient who had an ameloblastoma on her right mandible underwent segmental bone resection surgery. Then a customized titanium plate was inserted to the defect site. Plate fracture occurred in several months after surgery. To study the failure of the reconstruction plate, a numerical simulation using finite element method under different static chewing loadings was performed to calculate von Mises stress distribution in the plate and to determine the position of high stress concentration.

A reconstruction plate for repairing a mandible with continuity defects must reconstruct the bone shape anatomically. The external shape of the plate directly affects the recovery of facial appearance. Stock plates must be bended to adjust the plate shape to the mandible morphology for suitable placement, while customized plates with specific shape don't require intraoperative bending and offer better fitting.

In addition, the reconstruction plate must withstand the repeated forces created during mastication as well. In our simulations, the maximum von Mises stress in the plate was 260.55 MPa under incisor clenching, decreasing to 92.91 MPa under left molar clenching. These values are lower than the material failure limits of Ti6Al4 V (yield strength: 800 MPa and fatigue strength: 600 MPa) [4]. The maximum von Mises stress in the plate was observed on the inner-upper edge near the ramus end under both biting loads. Previously published literature found that the origin of cracks was stress concentration regions in the plate and stress concentration during repeated mastication was the origin of cracks, and it finally resulted in fatigue fracture failure. Although the calculated peak stress in the plate was below the failure limit of Ti6Al4 V, plate failure occurred. This might be explained by the existence of manufacturing defects in the plate which reduce its strength. The position of the plate fracture (Fig. 4) was in agreement with that of peak stress in the plate (Fig. 3). The result supports the reported findings that plate failure tends to occur at the stress concentration region. To prevent plate fracture, it is necessary to redesign the morphology of the plate for avoiding stress concentration. Therefore, smoothing the inner side of the plate and increasing heights and widths in the critical region may be helpful.

After segmental mandible resection surgery, the occlusal force was reduced in some degree. Muscle forces also changed due to bone resection and the remove of relevant tissue. There is still lack of information about muscle forces in patients with extensive mandibular defect. Previous work suggested that objective function-based numerical models provided a method to study muscle and joint forces for the individual. In this study, although the calculated muscle forces are exceedingly difficult to validate, it is useful to compare the plate biomechanical behavior under loads that it may be subjected to.

5 Conclusions

In conclusion, the finite element simulative results indicated that changing bite from molar region to incisor region increased both the maximum stress in the reconstruction plate and the maximum strain in the bone. It is suggested that the patient with lateral mandibular defects should reduce incisor clenching frequency. Additionally, stress concentration regions in the plate during mastication were likely regions of fracture failure. These critical regions should be strengthened in height and thickness for avoiding stress concentration.

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