Original Article

Craniomaxillofacial Trauma & Reconstruction

Craniomaxillofacial Trauma & Reconstruction
2021, Vol. 14(2) 135-143
© The Author(s) 2020
Article reuse guidelines:
sagepub.com/journals-permissions
DOI: 10.1177/1943387520952684
journals.sagepub.com/home/cmt
SAGE

Biomechanical In Vitro Study on the Stability of Patient-Specific CAD/CAM Mandibular Reconstruction Plates: A Comparison Between Selective Laser Melted, Milled, and Hand-Bent Plates

Robin Kasper, DMD[1], Karsten Winter, PhD[2], Sebastian Pietzka, MD, DDS[1,3], Alexander Schramm, MD, DDS[1,3], and Frank Wilde, MD, DDS[1,3]

Abstract

Study Design: An experimental in vitro study.

Objective: Plate fractures are a recurrent problem in alloplastic mandibular reconstruction. Hypothetically it can be assumed that computer-aided design (CAD)/computer-aided manufacturing (CAM) reconstruction plates have a higher stability than conventional hand-bent plates. The aim of the study was to compare additive and subtractive fabricated CAD/CAM mandibular reconstruction plates as well as conventional plates with regard to their biomechanical properties.

Methods: In a chewing simulator, plates of 2 conventional locking plate systems and 2 CAD/CAM-fabricated plate systems were compared. The plates were loaded in a fatigue test. The maximum number of cycles until plate fracture and the plate stiffness were compared.

Results: While all conventional plates fractured at a maximum load between 150 and 210 N (Newton) after a number of cycles between 40 000 and 643 000, none of the CAD/CAM plates broke despite a nearly doubled load of 330 N and 2 million cycles. Both CAD/CAM systems proved to be significantly superior to the hand-bent plates. There was no difference between the 2 CAD/CAM systems.

Conclusions: Concerning the risk of plate fracture, patient-specific CAD/CAM reconstruction plates appear to have a significant advantage over conventional hand-bent plates in alloplastic mandibular reconstruction.

Keywords

mandibular reconstruction, CAD-CAM, patient-specific implants, biomechanics

Introduction

Tumors, inflammations, and also osteonecrosis can require alveolar or segmental resection of the mandible. The latter is always going along with a loss of continuity of the jaw. This lost continuity should be restored for both aesthetic and functional reasons.[1,2] Primary reconstruction of the defect with autologous bone is the preferred option and should be undertaken whenever possible.[3-6] Papers suggest that even elderly patients might benefit from microvascular bony reconstruction.[7,8] In an increasing number of patients, however, this is not feasible or highly risky due to various concomitant and underlying diseases, so that a solely alloplastic reconstruction, usually using a titanium reconstruction plate, remains the last choice.[9-11] Still, problems with purely alloplastic mandibular reconstruction continue to be hardware-

[1] Department of Oral and Maxillofacial Surgery, University Hospital Ulm, Ulm, Germany
[2] Institute of Anatomy, Medical Faculty, Leipzig University, Leipzig, Germany
[3] Department of Oral and Plastic Maxillofacial Surgery, Military Hospital Ulm, Ulm, Germany

Corresponding Author:
Robin Kasper, DMD, Department of Oral and Maxillofacial Surgery, University Hospital Ulm, Albert-Einstein-Allee 11, 89081 Ulm, Germany.
Email: box@robinkasper.de

associated complications such as plate fractures or loosening of screws.[4,11-18] An important reason for this seem to be microcracks in the plate, which are produced during contouring with the aid of bending tools.[19,20]

The establishing and constant further development of computer-assisted surgery in cranio-maxillo-facial surgery and the new possibilities manufacturing patient-specific implants on the basis of computer-aided design (CAD)/computer-aided manufacturing (CAM) procedures lead to ever new therapeutic approaches.[21-23] Therefore, the idea of using patient-specific reconstruction plates manufactured by CAD/CAM procedures for purely alloplastic mandibular reconstruction seemed reasonable. Also in bony autologous reconstruction, they might play a vital role when the graft is not yet completely healed and therewith bearing the forces acting on it. Besides it is already known that patient-specific reconstruction plates offer significantly better results in terms of operating time, plate exposure, and occlusion.[22,24] These plates are either manufactured subtractively by computer-controlled milling from a titanium block or additively by means of selective laser melting in the sense of 3-dimensional printing. Due to these computer-assisted manufacturing processes, negative changes in metal structure, which occur when bending conventional plates,[19,20] are not to be expected. Furthermore, no special notches in the sense of predetermined bending points are required, which are necessary to enable the conventional plates to be contoured to the bone. It can therefore be assumed that CAD/CAM reconstruction plates have significantly increased stability and rigidity compared to conventional plates bent by hand. However, the extent of such a possible increase in stability is not yet known. Neither is there any information available on the extent to which the biomechanical properties of plates milled subtractively out of a titanium body differ from those plates additively produced by selective laser melting.

The aim of this in vitro study was therefore to compare both additive and subtractive manufactured CAD/CAM mandibular reconstruction plates and conventional plates in a fatigue test with regard to their biomechanical properties. The focus was put on the fatigue behavior up to plate fracture and the elastic deformation of the plates.

Material and Methods

This study features no human or animal subjects. No ethical approval was necessary because no biological materials or patient data were used in this in vitro study.

A standardized jaw model made of polyamide was used for the construction of the experimental model. The shape of the model was derived from the mandibular model "8311 Mandible Teeth Clip" by Synbone. For this purpose, the Synbone model was scanned using multislice computed tomography, reconstructed in 1 mm layering and sintered from polyamide by selective laser sintering after segmentation. The selected resection margins extended from the right mandibular angle to the premolar region of the left corpus.

Four different plate systems, 2 conventional locking and 2 CAD/CAM plate systems, were tested (Figure 1):

- A: DePuy Synthes, MatrixMANDIBLE Reconstruction Plate, angled right, profile thickness 2.5 mm
- B: KLS Martin, ThreadLock TS Reconstruction Plate, angled right, profile thickness 3.0 mm
- C: DePuy Synthes, TruMatch/ProPlan CMF (subtractively milled), profile thickness 2.5 mm
- D: KLS Martin, UNIQOS (Additive Manufacturing: Selective Laser Melting), profile thickness 2.5 mm

The 2 CAD/CAM plate systems (groups C and D) were manufactured following exactly the same design in analogy to the conventional plate systems (groups A and B). All 4 systems had a locking mechanism. A test number of $N = 10$ was determined for each of the 4 plate types.

The preparation of the test models, which included the contouring of the hand-bent plates and the following model operation, was performed for each system according to the manufacturer's recommendations. All plates were fixed with 4 screws each to the right mandibular ramus and 5 screws to the left corpus (Figure 1). Bicortical locking screws were used for all plates. The screw diameters range from 2.4 mm for DePuy Synthes MatrixMANDIBLE plates (group A), 2.7 mm for the KLS Martin ThreadLock plates (group B), 2.4 mm for the DePuy Synthes TruMatch system (group C) to 2.3 mm for the KLS Martin UNIQOS system (group D). To describe the screw localizations, the segments of the hand-bent plates were labeled and the screw holes were numbered (Figure 2).

The servo-hydraulic materials testing machine Amsler HC 10 from Zwick & Roell and the Workshop 96 Toolkit user program (Zwick & Roell) were used to perform the load tests. The plates were loaded in axial direction by a force actuator. In the oscillograph, this was displayed as a sinus curve in relation to time. The point of force application was located directly on the plate in the region of the first right lower premolar of the resected jaw section, which was intended to simulate both biting processes in the anterior region and grinding movements in the posterior region. The experimental model was positioned in the testing machine close to the physiological conditions with a clamp fixation at the condyles and a supporting point in the area of the mandibular angle. The suspension device with the model was mounted on a load cell (Figure 3).

A constant amplitude of ± 60 N around a varying mean load was chosen for the test, which, starting at 90 N, was successively increased by 20 N every 200 000 load cycles, unless plate failure occurred. Loading phases with the same mean load were called "force levels." The frequency was set continuously at 5 Hz (Hertz). The loading program ran in force-controlled mode. This means that overloading of the models was prevented by means of defined load limits.

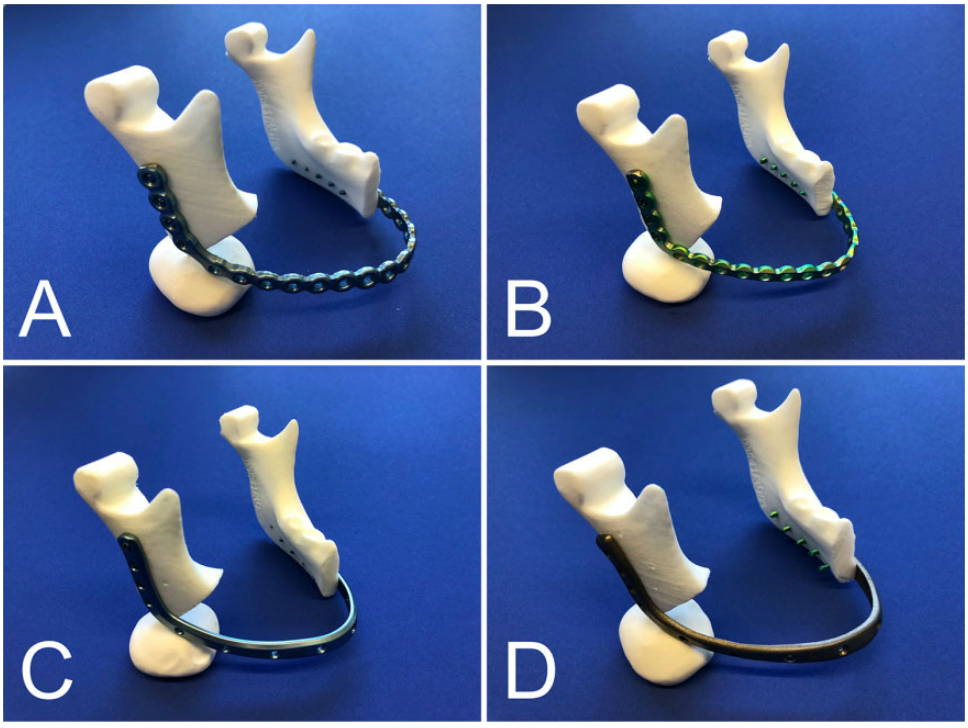


Figure 1. Prepared experimental models of the tested plate systems: (A) DePuy Synthes MatrixMANDIBLE, (B) KLS Martin ThreadLock, (C) DePuy Synthes TruMatch, and (D) KLS Martin UNIQOS (N = 10 each).

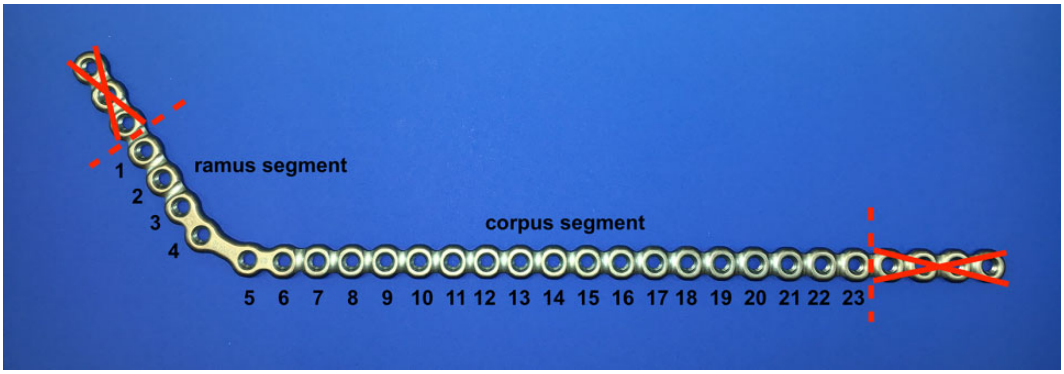


Figure 2. Labeling of the plate sections using a DePuy Synthes MatrixMANDIBLE plate (group A) as an example. The dotted lines mark the shortening points and the crosses the removed plate sections. The holes were numbered starting from the shortened ramus segment. The KLS Martin ThreadLock plates (group B) have 6 holes in the ramus segment and 20 in the corpus segment and have been shortened and labeled analogously to 4 holes in the ramus and 19 holes in the corpus segment. The screws were numbered in the same way as the holes.

In addition, displacement limits were defined to detect plate failure. This was defined as the point at which the plate exceeded the average maximum displacement during the current force level by 1.5 mm due to crack development or fracture.

The programs SPSS (Version 25; IBM) and R (Version 3.6.1; R Core Team; https://www.R-project.org) were used for statistical analysis. The recorded maximum number of cycles until failure of the plates was tested for normal distribution within each plate system using a Kolmogorov-Smirnov test and Shapiro-Wilk test. The following significance analysis was done by Mann-Whitney U test and a defined significance level of $\alpha = .05$.

To assess the stiffness of the plates, the maximum vertical displacement during the maximum load acting on them was observed. Mean values were calculated for each test plate within a force level, which in turn were averaged within a plate system. In order to ensure the best possible

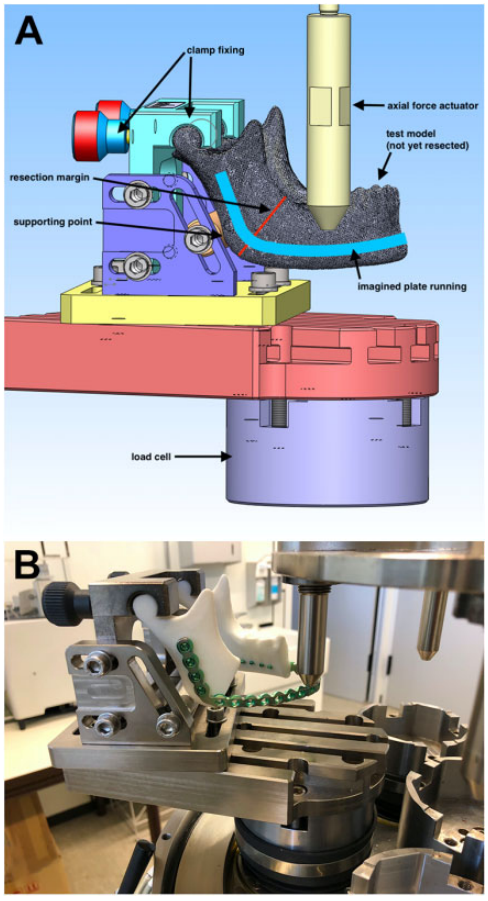


Figure 3. A, Schematic picture of the experimental setup with (not yet resected) jaw model. B, Finished trial setup.

validity of the mean values, the presence of a mean maximum displacement value of at least 3 plates was demanded for each plate system. Then a straight-line equation was calculated for each plate system using linear interpolation. These equations include the slope and *y*-axis intersection of the particular straight line.

Next, the averaged values of all plate systems were transferred into a linear regression model. The linear regression was performed using the least squares method (generalized least squares), since the data underlie autocorrelation (series of measurements over a time span) and heteroskedasticity (different scattering in the course of the measurement) could possibly be present. To investigate the latter, the Breusch-Pagan test was used. Now the test series were compared with each other in terms of slope and intersection with the *y*-axis using the regression model. Both values were thus tested for significant differences between the different plate systems. Due to multiple tests, the previously determined *P* value of .05 was adjusted with the help of a post hoc test. In this way, the Tukey's test was used to additionally test the individual plate systems pairwise for significant differences between each other.

Table 1. Number of Load Cycles to Plate Failure Within Groups A (DePuy Synthes MatrixMANDIBLE) and B (KLS Martin ThreadLock).[a]

Load cycles until plate fracture	(Group A) DePuy Synthes MatrixMANDIBLE	(Group B) KLS Martin ThreadLock
Median	375 409.50	55 263.50
Mean	347 330.10	71 601.00
Standard deviation	158 124.82	49 442.98

[a]The median, the mean, and the standard deviation are shown.

Results

Maximum Number of Load Cycles (Continuous Load Capacity)

None of the 20 CAD/CAM plates of the groups C (DePuy Synthes TruMatch) and D (KLS Martin UNI-QOS) failed over the entire range of 2 million cycles up to a maximum load of 330 N. There were no fractures or loosening of the screws. A significant difference was found between the hand-bent plate systems (groups A and B) in terms of long-term loading capacity ($P < .05$). On average, group A (DePuy Synthes MatrixMANDIBLE; mean value = 347 330.10 load cycles) achieved more than 4.8 times the number of cycles of the group B (KLS Martin ThreadLock; mean value = 71 601 load cycles) (Table 1). Due to the non-failure of the CAD/CAM plates (groups C and D), no maximum number of cycles could be determined, so that a statistical analysis could not be performed.

The fracture localizations of the conventional hand-bent plates differed both between the plate systems and partly within the system itself and are shown in Table 2. Figure 4 shows a typical fracture localization as it occurred in group A (DePuy Synthes MatrixMANDIBLE). Two test plates of group B (KLS Martin ThreadLock) showed loosening of the screw #19.

Elastic Deformation Behavior (Stiffness)

As only test plate number 2 of group B (KLS Martin ThreadLock) was able to reach the second force level, it was not possible to determine reliable mean values for this type of plate, so that this series had to be excluded from the significance analysis. In addition, only mean values for the first 3 force levels could be calculated for group A (DePuy Synthes MatrixMANDIBLE), since only 1 plate reached the fourth force level.

With regard to both the straight-line slope and the point of intersection with the *y*-axis, group A (DePuy Synthes MatrixMANDIBLE) differs significantly from both CAD/CAM systems ($P < .001$ each). The 2 CAD/CAM reconstruction systems (groups C and D) showed

Table 2. Overview of Fracture Localizations of the Individual Test Plates (N = 10 each) in Groups A (DePuy Synthes MatrixMANDIBLE) and B (KLS Martin ThreadLock).

Plate no.	(Group A) DePuy Synthes MatrixMANDIBLE	(Group B) KLS Martin ThreadLock
1	Between screw holes 18 and 19	Between screw holes 3 and 4
2	Between screw holes 18 and 19	Between screw holes 3 and 4
3	Between screw holes 11 and 12	Between screw holes 3 and 4
4	Between screw holes 12 and 13	Between screw holes 3 and 4
5	Between screw holes 6 and 7	Between screw holes 3 and 4
6	Between screw holes 19 and 20	Between screw holes 3 and 4
7	Between screw holes 18 and 19	Between screw holes 3 and 4
8	Between screw holes 18 and 19	Between screw holes 3 and 4
9	Between screw holes 11 and 12	Between screw holes 3 and 4
10	Between screw holes 18 and 19	Between screw holes 3 and 4



Figure 4. Fracture of a DePuy Synthes MatrixMANDIBLE plate (group A). The picture shows a typical fracture location proximate to the resection margin (between screw holes 18 and 19).

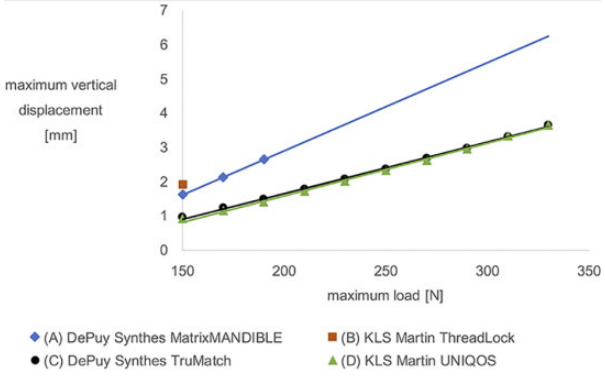


Figure 5. The 3 determined straight lines are shown. They visualize the elastic deformation behavior of the plate systems, that is, the deformation in mm (maximum vertical displacement) as a function of the maximum load in Newton (N). As explained in the text, no straight-line equation could be determined for group B (KLS Martin ThreadLock). Thus, for the complete picture the mean value of the first force level (150 N) of group B is displayed.

no significant difference between each other ($P = .490$ and $P = .253$). Thus, both CAD/CAM systems have a significantly higher stiffness than the 2 hand-bent plate systems (groups A and B). The straight lines and the results of the significance analysis are shown in Figure 5 and Table 3.

Table 3. Results of the Group Analyses.

	Difference (slope of the line)	Difference (intersection with *y*-axis)
(Group A) DePuy Synthes MatrixMANDIBLE versus (group C) DePuy Synthes TruMatch	Significant ($P < .001$)	Significant ($P < .001$)
(Group A) DePuy Synthes MatrixMANDIBLE versus (group D) KLS Martin UNIQOS	Significant ($P < .001$)	Significant ($P < .001$)
(Group C) DePuy Synthes TruMatch versus (group D) KLS Martin UNIQOS	Not significant ($P = .490$)	Not significant ($P = .253$)

[a]A total of 6 tests were performed. Series A (DePuy Synthes MatrixMANDIBLE) against C (DePuy Synthes TruMatch), A (DePuy Synthes MatrixMANDIBLE) against D (KLS Martin UNIQOS), and C (DePuy Synthes TruMatch) against D (KLS Martin UNIQOS) were tested with respect to the slope of the straight line and the point of intersection with the *y*-axis. *P* (= *P* value) is the exceedance probability. The significance level was set at $\alpha = .05$ and adjusted using the Tukey's test.

Discussion

Taking into consideration that the experimental design aims to approach the physiological conditions of the human body as closely as possible, the construction of the test model plays a major role. Human mandibles represent ideal material properties, but they are difficult to store and carry a risk of infection. Additionally, the specimens differ in terms of their anatomy, which makes comparability difficult. Synthetic mandibular models are cheap, always have the same design, and can be simplified at certain areas, which makes the experimental setup easier. The choice of a polyamide model with a standardized condyle design ensured the comparability and reproducibility of the tests and reduced the number of samples required.

A critical point to note is that only 1 defect class was tested. However, since the aim was to compare the different plate systems against each other and not to determine parameters directly applicable to the body, the selection of a single defect class seemed sufficient for this purpose.

Lateral defects in the molar and premolar region not only are the most frequent defect class but are also predisposed to hardware-associated complications.[15,17,25-27] In addition, increased complication rates in general can be found for defects that cross the midline.[4,28] Most in vitro and also finite element method (FEM) studies chose unilateral defects and less studies focused on defects crossing the midline. Even fewer of these include patient-specific reconstruction plates.[29] Since the defect class seems to be particularly relevant for alloplastic reconstruction and in order to extend the lack of available data in this respect, we have decided to further concentrate on this defect class.

When simulating a physiological loading, such an in vitro experiment reaches the limits of what is possible. The movements of the mandible are characterized by complex motions in all 3 dimensions. The lower jaw follows masticatory muscle forces in forms of sagittal and transversal bending and torsion of the corpus.[30,31] After a resection of the mandible, these principles are no longer applicable, not least because of the often asymmetrical loading of the remaining dentulous jaw. Moreover, the defect chosen in this study represents a major discontinuity, which in reality often leads to patients being unable to chew on their remaining teeth at all or only to a very limited extent. In addition, this may result in new adaptive jaw movements with new stress foci.[32] It seems hardly feasible to integrate these circumstances into the experimental setup. In addition to the microcracks of the hand-bent plates mentioned in the introduction, a further discussed factor influencing the increased fracture susceptibility of reconstruction plates after alloplastic reconstruction is the loss of the proprioceptive feedback of the masticatory system. The radical surgery results in the loss of essential structures, which might lead to an excessive overactivity of the masticatory muscles and thus increased stress on the plate. This makes it even more difficult to construct an in vitro model that incorporates this pathophysiology.[33,34] Since it was not our aim to restore the exact biomechanical setting and evaluate its effect on the plates, we chose a more simple and predictable setup to focus on the comparison of the plates regarding their long-term stability and stiffness. Therefore it seemed adequate for us using a purely axial load while allowing the model to perform torsional movements through the plain support in the area of the mandibular angle.

The determination of the loading force was based on the current data regarding the maximum human bite force. In the literature, sometimes widely divergent data were found. The maximum bite force in people with full dentition varies between 242.4 and 539.5 N or 31 and 443 kg depending on gender, age, and measuring technique. The average can be stated to be approximately 420 N.[35-42] With regard to the bite force of the resected jaw, far less information is found. Maurer et al state a reduction of 76% in the molar region and 59% in the incisal region.[43] The determination of the maximum load of 330 N in the premolar region in the present study therefore seemed to come close to reality. Nevertheless, the variances of the data given in the literature, individual differences, and other influencing factors like the type of defect must not be disregarded when interpreting the results.

None of the CAD/CAM plates showed signs of failure, which may lead to the hypothesis that the CAD/CAM plates do not break in the clinical use. However, due to the much higher number of variables in vivo, it must be pointed out that failure of these patient-specific CAD/CAM plates can never be ruled out, which is meanwhile also shown in the clinical setting.[44] Nevertheless, the results are very promising and may be a treatment alternative to hand-bent plates for certain patients. These could be patients in whom immediate or early defect bridging with an autologous bone graft is not or no longer possible, or in whom bony reconstruction does not contribute to an improvement in quality of life. Among them are often old, multimorbid patients as well as patients with advanced cancer, for whom palliative therapy is the preferred option and additional operations should be avoided. Furthermore, such CAD/CAM plates could possibly be suitable for patients with drug-induced osteonecrosis of the jaw, who are predominantly in a palliative situation with limited life expectancy. This could reduce the risk of an early plate replacement due to a fracture or avoid further surgery for bony reconstruction.

The plate diameter of group B (KLS Martin Thread-Lock) being 3 mm differs from the other groups with 2.5 mm in thickness. Out of their product range, KLS Martin recommended this system for the type of alloplastic reconstruction that was performed in this study. In our opinion, it was reasonable to use these plates, because especially due to different material properties and manufacturing processes the biomechanics including fatigue behavior are not only dependent on the diameter. Certainly, the results cannot be transferred to the plate diameter alone. However, the plates are comparable in that they share the same indication. Completely identical plates from 2 manufacturers are unlikely to be found, since, as mentioned above, there are many factors that influence biomechanics. Thus, the results can rather be landmarks for further studies. Looking at the continuous load capacity of the 2 hand-bent plate systems, it seems remarkable that group B, despite its thicker diameter of 3 mm, achieved significantly lower maximum cycle numbers than group A (DePuy Synthes MatrixMANDIBLE) with a plate diameter of 2.5 mm. Another difference can be seen in the localization of the fractures. While the plates of group A fractured primarily in the area of the screw closest to the resection margin (between holes 18 and 19), all plates of group B showed their failure in the junction area between screws 3 and 4 in the mandibular angle. A plausible explanation could be subjectively perceived higher bending stresses in the area of the mandibular angle necessary to contour the plates. Larger diameters require higher forces in order to be bended. These higher

forces in turn could lead to greater damage to the plate surface by the bending tools and to the internal metal structure, which would be associated with an increased tendency to fracture. Gutwald et al describe a similar fracture localization in a comparable in vitro study.[45]

The fracture localization found in half of group A (prior to the first screw distal to the resection margin of the stump contralateral to the loaded side) is also described as a predisposed site for plate fractures. Bujtár et al showed with the help of comparable FEM models that the highest forces during masticatory loading are concentrated around this screw.[46] In addition, 2 plates of group B showed a loosening of this screw (here no. 19). Exact reasons for this were not apparent. One possible influence could be the larger screw diameter of 2.7 mm compared to groups A, C, and D, which measure 2.4, 2.4, and 2.3 mm, respectively. One study that contradicts this hypothesis was the one of Gateno et al. They compared the dislocation of 2.4 and 2.0 mm diameter screws in a very similar stress test. No significant differences were found. Screw fractures did not occur.[47] Another argument against this approach would be the consideration that the larger surface area increases the friction between bone and screw, which in turn leads to greater insertion force and thus to a potentially tighter fit of the screw. Other possible reasons could be material defects of the screw, plate, or jaw model, incorrect insertion, or the material properties of the polyamide model. The records did not show any abrupt changes in displacement, so that a slow development can be assumed. Whether and to what extent this loosening influenced the fracture susceptibility of the plate could not be conclusively clarified.

The significantly increased continuous loading capacity of the CAD/CAM plates and the associated conceivable indications already described may bring about some new factors that need to be considered. Due to the longer remaining of the plate in situ and potentially increasing loads, the forces not only on the plate but also on the screws increase, which, in contrast to the plate, do not have an increased continuous loading capacity. As described above, these forces are highest in the area of the screw closest to the resection margin. It therefore seems advisable to fix patient-specific CAD/CAM plates, as tested here, with a sufficient number of screws per residual stump in order to reduce the risk of screw fracture up to a potential screw tear out of the bone.

The fatigue strength of metallic materials depends to a large extent on the alloy used. All plates tested here have titanium as their main component. The following metals were used by the companies: groups A and C (DePuy Synthes MatrixMANDIBLE and DePuy Synthes TruMatch)—pure titanium (cpTi4 = commercially pure titanium, grade 4 according to ASTM F67 and ISO 5832-2); group B (KLS Martin ThreadLock)—pure titanium (cpTi2, grade 2 according to ASTM B265-15); group D (KLS Martin UNIQOS)—titanium alloy (TiAl6V4 according to ASTM F136-02a). TiAl6V4 is a titanium alloy that, in addition to titanium, contains about 6 mass percent aluminum and 4 mass percent vanadium. According to Zimmer, the fatigue strength of annealed pure titanium over 107 cycles is 241 to 379 MPa whereas that of annealed TiAl6V4 alloy is 379 to 448 MPa.[48] This means that the plates of group D (KLS Martin UNIQOS) have a clear advantage. Looking at the results, however, the alloy does not seem to play a significant role, at least not in the plates used in this experiment. Nonetheless, there are a number of other factors that influence the fatigue behavior, making it almost impossible to predict it. This concerns particularly the type of manufacturing. In selective laser melting, the material, particle size, homogeneity, and contamination of the metal powder used, as well as the power and beam diameter of the laser tool, the scanning process, the parameters of the entire system, the layer thickness, and the direction in which the structure is built up play an important role.[49] Further influencing factors are the plate geometry, the type of load,[50,51] and, in the case of hand-bent plates, the aforementioned vulnerability to bending stress.

In summary, it is eventually up to in vitro studies like this one and the clinical experience to compare and classify the variety of medical implants offered in terms of their biomechanical value.

According to the results on elastic deformation behavior, the CAD/CAM plates have a significantly higher stiffness than the hand-bent ones. It is now important to know whether the load on the individual screw changes with increasing stiffness of the plate and if so, whether this in turn leads to a more even distribution of the forces on the screws. This in turn would have the consequence that the risk of screw tearing would decrease due to the lower forces on the individual screws. However, this assumption would be contradicted by the longer average loading time or loading force of patient-specific plates. Thus, it remains to be examined in further studies what effects the changed material properties of CAD/CAM reconstruction plates have on the biomechanics of the entire system of alloplastic mandibular reconstruction.

Conclusions

In terms of reducing the risk of plate fracture in vivo, patient-specific CAD/CAM reconstruction plates appear to have a significant advantage over conventional hand-bent plates and could therefore be a valuable alternative in cases where alloplastic reconstruction cannot be avoided. These could, for example, be multimorbid patients or younger patients with higher expected chewing forces. Even in cases of osteocutaneous reconstruction, the plates offer a good immediate stability until the transplant is healed.

Availability of Data and Material

The data sets generated and analyzed during the current study are available from the corresponding author on reasonable request.

Declaration of Conflicting Interests

The author(s) declared no potential conflicts of interest with respect to the research, authorship, and/or publication of this article.

Funding

The author(s) received no financial support for the research, authorship, and/or publication of this article: The project was funded by the special research project of the German Armed Forces with project number 35K3-S-81 1517.

References

1. Baker A, McMahon J, Parmar S. Immediate reconstruction of continuity defects of the mandible after tumor surgery. *J Oral Maxillofac Surg*. 2001;59(11):1333-1339.
2. Linsen S, Schmidt-Beer U, Grüner M, Koeck B. CMD-Beschwerden bei Patienten mit Defekten im Kieferbereich. *Dtsch Zahnaerztl Z*. 2005;60:376-387.
3. Bianchi B, Ferri A, Ferrari S, et al. Reconstruction of lateral through and through oro-mandibular defects following oncological resections. *Microsurgery*. 2010;30(7):517-525.
4. van Gemert JTM, van Es RJJ, Rosenberg AJWP, van der Bilt A, Koole R, van Cann EM. Free vascularized flaps for reconstruction of the mandible: complications, success, and dental rehabilitation. *J Oral Maxillofac Surg*. 2012;70(7):1692-1698.
5. Urken ML, Buchbinder D, Weinberg H, et al. Functional evaluation following microvascular oromandibular reconstruction of the oral cancer patient: a comparative study of reconstructed and nonreconstructed patients. *Laryngoscope*. 2015;125(7):1512.
6. Kumar BP, Venkatesh V, Kumar KAJ, Yadav BY, Mohan SR. Mandibular reconstruction: overview. *J Maxillofac Oral Surg*. 2016;15(4):425-441.
7. Torabi SJ, Chouairi F, Dinis J, Alperovich M. Impact of advanced age on microvascular reconstruction of the lower facial third: an American College of Surgeons NSQIP study. *Microsurgery*. 2019;39(6):487-496.
8. Sugiura Y, Sarukawa S, Hayasaka J, Kamochi H, Noguchi T, Mori Y. Mandibular reconstruction with free fibula flaps in the elderly: a retrospective evaluation. *Int J Oral Maxillofac Surg*. 2018;47(8):983-989.
9. Head C, Alam D, Sercarz JA, et al. Microvascular flap reconstruction of the mandible: a comparison of bone grafts and bridging plates for restoration of mandibular continuity. *Otolaryngol Head Neck Surg*. 2003;129(1):48-54.
10. Wei F, Celik N, Yang W, Chen I, Chang Y, Chen H. Complications after reconstruction by plate and soft-tissue free flap in composite mandibular defects and secondary salvage reconstruction with osteocutaneous flap. *Plast Reconstr Surg*. 2003;112(1):37-42.
11. Dannemann S, Abu-Id MH, Kreusch T. Die zweizeitige Unterkieferrekonstruktion mit einer 2,7-mm-Überbrückungsplatte. *Mund Kiefer Gesichtschir*. 2007;11(6):339-343.
12. Freitag V, Hell B, Fischer H. Experience with AO reconstruction plates after partial mandibular resection involving its continuity. *J Craniomaxillofac Surg*. 1991;19(5):191-198.
13. Irish JC, Gullane PJ, Gilbert RW, Brown DH, Birt BD, Boyd JB. Primary mandibular reconstruction with the titanium hollow screw reconstruction plate: evaluation of 51 cases. *Plast Reconstr Surg*. 1995;96(1):93-99.
14. Arden RL, Rachel JD, Marks SC, Dang K. Volume-length impact of lateral jaw resections on complication rates. *Arch Otolaryngol Head Neck Surg*. 1999;125(1):68-72.
15. Coletti DP, Ord R, Liu X. Mandibular reconstruction and second-generation locking reconstruction plates: outcome of 110 patients. *Int J Oral Maxillofac Surg*. 2009;38(9):960-963.
16. Maurer P, Eckert AW, Kriwalsky MS, Schubert J. Scope and limitations of methods of mandibular reconstruction: a long-term follow-up. *Br J Oral Maxillofac Surg*. 2010;48(2):100-104.
17. Liu S, Cai Z, Zhang J, Zhang J, Zhang Y. Plate related complication after mandibular reconstruction. *Zhonghua Kou Qiang Yi Xue Za Zhi*. 2012;48(10):586-590.
18. Chung J-H, Yoon E-S, Park S-H, Lee B-I, Kim H-S, You H-J. Comparison of mechanical stability between fibular free flap reconstruction versus locking mandibular reconstruction plate fixation. *Arch Craniofac Surg*. 2014;15(2):75-81.
19. Brunner H, Simpson JP. Fatigue fracture of bone plates. *Injury*. 1980;11(3):203-207.
20. Martola M, Lindqvist C, Hänninen H, Al-Sukhun J. Fracture of titanium plates used for mandibular reconstruction following ablative tumor surgery. *J Biomed Mater Res B Appl Biomater*. 2007;80(2):345-352.
21. Wilde F, Cornelius C-P, Schramm A. Computer-assisted mandibular reconstruction using a patient-specific reconstruction plate fabricated with computer-aided design and manufacturing techniques. *Craniomaxillofac Trauma Reconstr*. 2014;7(2):158-166.
22. Wilde F, Hanken H, Probst F, Schramm A, Heiland M, Cornelius C-P. Multicenter study on the use of patient-specific CAD/CAM reconstruction plates for mandibular reconstruction. *Int J Comput Assist Radiol Surg*. 2015;10(12):2035-2051.
23. Mascha F, Winter K, Pietzka S, Heufelder M, Schramm A, Wilde F. Accuracy of computer-assisted mandibular reconstructions using patient-specific implants in combination with CAD/CAM fabricated transfer keys. *J Craniomaxillofac Surg*. 2017;45(11):1884-1897.
24. Sieira Gil R, Roig AM, Obispo CA, Morla A, Pagès CM, Perez JL. Surgical planning and microvascular reconstruction of the mandible with a fibular flap using computer-aided design, rapid prototype modelling, and precontoured titanium reconstruction plates: a prospective study. *Br J Oral Maxillofac Surg*. 2015;53(1):49-53.
25. Shibahara T, Noma H, Furuya Y, Takaki R. Fracture of mandibular reconstruction plates used after tumor resection. *J Oral Maxillofac Surg*. 2002;60(2):182-185.

26. Katakura A, Shibahara T, Noma H, Yoshinari M. Material analysis of AO plate fracture cases. *J Oral Maxillofac Surg*. 2004;62(3):348-352.
27. Ettl T, Driemel O, Dresp BV, Reichert TE, Reuther J, Pistner H. Feasibility of alloplastic mandibular reconstruction in patients following removal of oral squamous cell carcinoma. *J Craniomaxillofac Surg*. 2010;38(5):350-354.
28. Markwardt J, Pfeifer G, Eckelt U, Reitemeier B. Analysis of complications after reconstruction of bone defects involving complete mandibular resection using finite element modelling. *Onkologie*. 2007;30(3):121-126.
29. Rendenbach C, Sellenschloh K, Gerbig L, et al. CAD-CAM plates versus conventional fixation plates for primary mandibular reconstruction: a biomechanical in vitro analysis. *J Craniomaxillofac Surg*. 2017;45(11):1878-1883.
30. Koolstra JH. Dynamics of the human masticatory system. *Crit Rev Oral Biol Med*. 2002;13(4):366-376.
31. van Eijden TM. Biomechanics of the mandible. *Crit Rev Oral Biol Med*. 2000;11(1):123-136.
32. Atkinson HF, Shepherd RW. The masticatory movements of patients after major oral surgery. *J Prosthet Dent*. 1969;21(1): 86-91.
33. Cantor R, Curtis TA. Prosthetic management of edentulous mandibulectomy patients. I. Anatomic, physiologic, and psychologic considerations. *J Prosthet Dent*. 1971;25(4): 446-457.
34. Schaaf NG. Oral reconstruction for edentulous patients after partial mandibulectomies. *J Prosthet Dent*. 1976;36(3): 292-297.
35. Helkimo E, Carlsson GE, Helkimo M. Bite force and state of dentition. *Acta Odontol Scand*. 1977;35(6):297-303.
36. Sonnenburg M, Fethke K, Riedel S, Voelker H. Zur Belastung der Zähne des menschlichen Kiefers. *Zahn Mund Kieferheilkd Zentralbl*. 1978;66(2):125-132.
37. Gibbs CH, Mahan PE, Lundeen HC, Brehnan K, Walsh EK, Holbrook WB. Occlusal forces during chewing and swallowing as measured by sound transmission. *J Prosthet Dent*. 1981;46(4):443-449.
38. Proffit W, Fields H, Nixon W. Occlusal forces in normal- and long-face adults. *J Dent Res*. 1983;62(5):566-570.
39. Gibbs CH, Mahan PE, Mauderli A, Lundeen HC, Walsh EK. Limits of human bite strength. *J Prosthet Dent*. 1986;56(2): 226-229.
40. Kampe T, Haraldson T, Hannerz H, Carlsson GE. Occlusal perception and bite force in young subjects with and without dental fillings. *Acta Odontol Scand*. 1987;45(2):101-107.
41. Lauer H-Ch, Ottl P, Ficker E, Weigl P. Kaukraftmessung an unterschiedlich verankerten Teilprothesen. *Dtsch Zahnaerztl Z*. 1992;47:482-485.
42. Harada K, Watanabe M, Ohkura K, Enomoto S. Measure of bite force and occlusal contact area before and after bilateral sagittal split ramus osteotomy of the mandible using a new pressure-sensitive device: a preliminary report. *J Oral Maxillofac Surg*. 2000;58(4):370-374.
43. Maurer P, Pistner H, Schubert J. Computer assisted chewing power in patients with segmental resection of the mandible. *Mund Kiefer Gesichtschir*. 2006;10(1):37-41.
44. Telschow T, Wilde F, Pietzka A, Schramm A, Mascha F. Unbreakable?—Frakturen patientenspezifischer Implantate nach alloplastischer Unterkieferrekonstruktion. *Der MKG-Chirurg*. 2019;12:263-267.
45. Gutwald R, Jaeger R, Lambers FM. Customized mandibular reconstruction plates improve mechanical performance in a mandibular reconstruction model. *Comput Methods Biomech Biomed Engin*. 2017;20(4):426-435.
46. Bujtár P, Simonovics J, Váradi KB, Sándor GKB, Avery CME. The biomechanical aspects of reconstruction for segmental defects of the mandible: a finite element study to assess the optimization of plate and screw factors. *J Craniomaxillofac Surg*. 2014;42(6):855-862.
47. Gateno J, Cookston C, Hsu SS, et al. Biomechanical evaluation of a new MatrixMandible plating system on cadaver mandibles. *J Oral Maxillofac Surg*. 2013;71(11):1900-1914.
48. Zimmer USA. *Physical and Mechanical Properties of Orthopaedic Alloys. Zimmer Technical Monograph*. Elsevier; 1984.
49. Zerbst U, Hilgenberg K. Schadensentwicklung und Schadenstoleranz von SLM-gefertigten Strukturen. In: Richard H, Schramm B, Zipsner T, eds. *Additive Fertigung von Bauteilen und Strukturen*. Springer Vieweg; 2017:242.
50. Haug RH, Fattahi TT, Goltz M. A biomechanical evaluation of mandibular angle fracture plating techniques. *J Oral Maxillofac Surg*. 2001;59(10):1199-1210.
51. Doty JM, Pienkowski D, Goltz M, Haug RH, Valentino J, Arosarena OA. Biomechanical evaluation of fixation techniques for bridging segmental mandibular defects. *Arch Otolaryngol Head Neck Surg*. 2004;130(12):1388-1392.