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ORIGINAL ARTICLE

Finite element analysis of customized implant in mandibular reconstruction after tumor resection with and without using customized surgical osteotomy guide

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Abstract

Objectives: The aim of this work was to compare different 12 cases (3 patients * 4 cases = 12 cases) with varying gaps between implant and bone by analyzing the effect of these gaps on implant and screws using FEM.

Methods: In each patient's case 1 using CSOG and in case 2, 3, and 4 without using CSOG tumor cutting was done. Hence in each patient zero gaps at case 1 and overcutting at case 2, 3, and 4 have obtained at different locations.

Results: FEM results reveal that in each patient's case 4 (maximum gap) was more susceptible to loosening of the screws due to higher strains (37%) and implant failure due to higher stress (28%) concentration under the same loading conditions when compared with case 1 (zero gap).

Conclusions: The study reveals that mandibular reconstruction with implant placement using CSOG can significantly enhance the stability and safety of the implant.

KEYWORDS

ablative tumor surgery, customized mandibular implant, customized surgical osteotomy guide, finite element analysis, mandibular continuity defect

1 | INTRODUCTION

Ablative tumor therapy is one of the main important causes for mandibular continuity defects. Reconstruction surgery of continuity defects of the mandible is still challenging for craniomaxillofacial (CMF) surgeons.[1] This type of surgery becomes complex and unpredictable because of very limited visibility of closed internal structures, the presence of teeth and their relationship with bone, the influence of surgery on the airway, and interference with occlusion.[2] Mainly, different extended mandibular malignant and benign lesions and inflammation are treated using ablative surgery, which can cause continuity defects in the mandible.[1] Furthermore, squamous cell carcinomas (SCC) of the tongue, the floor of the mouth and mandibular alveolar process are treated with ablative tumor therapy.[1] Accurate mandibular reconstruction is needed for functional and aesthetic improvement. Loss of anatomical mandibular shape includes problems in speech, mastication, deglutition, and occlusion.[1]

In current practices, the most reliable therapy is an instant reconstruction of defects using musculo-osseous flaps which are microvascularly anastomosed and, harvested from the fibula, iliac crest, or scapula.[1,3] However, the instant reconstruction of continuity defects of the mandible is not always possible. These techniques are sensitive, associated with donor site morbidity, and may have a limitation in shape and size of the graft, which often precludes their use.[2]

One standard approach is prefabricated metallic reconstruction plates, which plays a vital role in bridging continuity defects and maintaining remaining stumps on accurate site in order to assure ingestion, speech, and patency of the upper airways after ablative tumor surgery of the mandibular region.[1,3] Clinical experience and previous literature have shown that these commercially available standard reconstruction plates are often subject to excessive stress that may lead to fatigue fractures.[4] These plates are designed for the 'average' patients and are supplied as straight or slightly contoured metal plates with preformed retention screw holes in only generic shapes and sizes.[5] During surgery, the surgeon may have to spend a significant amount of time in bending and shaping of the plate to fit the contour of the patient's bone.[2,6] They often fail to reproduce accurate anatomical structural shape. Bending of the plate might increase the chances of plate fracture from weak points.

Plate loosening, fracture, or outside exposure are common complications arising with reconstruction plates.[1,3,7] The use of standard plates in simple surgical procedures offers good results in straightforward cases. In complex cases, however, standard plates often lack passive anatomical fitting that can be overcome by using customized plates. But the disadvantage of the customized plate is that it cannot achieve shape similar to the resected bone.

To avoid this complication and to improve the results, pre-bending of plates on rapid prototyping (RP) models is a new technique.[3] But exact bending of the plate similar to the bone contour could not be obtained because of the complex shape of the mandible and uneven shape and size of the tumor. Due to the improper adaption of the plate, there may be chances of incongruity between plate and contour of the bone. This may create a dead gap between bone and surrounding soft tissues. Less soft tissue might produce high tension which leads to exposure of plate.[3] The plate stability was slightly improved by modifying plate shape and fixture systems. However, the rate of complications was not reduced significantly. Major disadvantage of this technique is it require diseased RP model for plate adaptation.

Nowadays fabrication of customized titanium trays combined with autologous bone is one technique to accomplish such implants.[8] By using this technique, mechanically stable bridging of continuity defects is possible. Custom made mesh trays consisting of raw particular hydroxyapatite and poly-l-lactide to restore the mandible is also another new method.[9] These methods require a highly skilled person for mesh tray design.

In order to avoid the above-mentioned complications, one promising approach seems to be the application of shape identical, functionally stable implants to reconstruct continuity defects of the mandible.[1] These customized implants conform to the external shape of the defect site (resected bone) that is intended to be replaced.[3,10] These customized implants could be developed preoperatively to fit exactly by tube-in-tube like connections to the remaining stumps. Thus customized implants prevent the intra-operative adaptation. Also, in craniofacial surgery these customized implants provide better fit and cosmesis, faster recovery, and reduces operating time as well as chances of infection.[10]

Generally, these customized implants are designed using patient's clinical image data, virtual surgical planning (VSP), and computer-aided design (CAD). At the time of design the surgeon finalized the tumor size, surgical margin, cutting locations, screw locations, and number of screws needed. Rapid prototyping offers preoperative fabrication of customized implants.[11] Numerous rapid prototyping techniques (EBM, SLS or DMLS) have been developed for the fabrication of such a highly complex individual shape implants in metal forms like titanium or cobalt chrome. Markwardt *et al*.[1] and Schoene *et al*.[12] used LaserCUSING® a laser sintering technology for manufacturing of customized implants.

The latest techniques fulfill the basic requirement of design and development of accurate customized implant. However, because of improper resection (overcutting) of the tumor, the implant may not precisely fit to the planned location. Due to overcutting, a huge gap between implant and bone may develop. Sometimes all planned screws may not fit on the planned location or some screws may enter into the gap. Because of improper fixing of the implant to the bone, high mechanical stresses and strains develop on implant, screws, and screw holes in the bone, which leads to screw loosening, implant failure and hence re-surgery.

Therefore, for mandibular reconstruction after ablative tumor surgery, precisely customized implant fixation at the preplanned location is very important. Exact tumor resection is an essential factor for accurate fixation of customized implant. This implant should exactly match with resected bone to achieve stable connection. To the best of our knowledge, effects of precise fixation of customized implants have not been given in any previous literature. This opens up questions regarding the structural strength of mandibular reconstruction after placement of customized implants and the effect of improper fixation of the implant due to overcutting, in terms of structural performance such as flexibility, strength, fixation, and the interaction between screws and bone tissues. A customized surgical osteotomy guide (CSOG) (Figure 1), is a device that plays a vital role for exact tumor resection and accurate implant fixation. This device is designed using patient's CT scan at the time of VSP and manufactured using rapid prototyping (RP). In VSP, as per the size of the tumor, the size of customized implant and CSOG are finalized. This warrants a holistic approach to the implants fixation after ablative tumor therapy, where biomechanics theory must be incorporated in reconstruction formalizing implant fixation paradigms.

The aim of this pilot study was to examine the hypothesis that a stable fixation of customized titanium implant between the mandibular

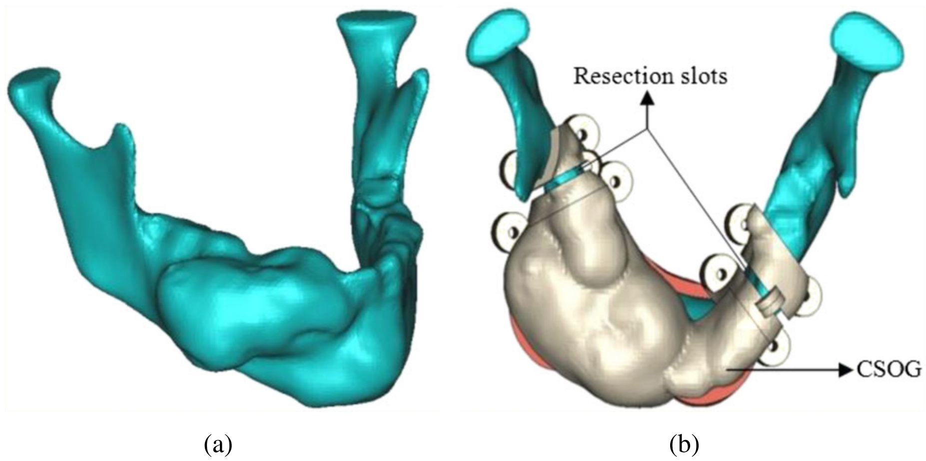

FIGURE 1 (a) 3D reconstruction of the diseased mandible of first patient, and (b) customized surgical osteotomy guide (CSOG) for accurate tumor resection

stumps could be achieved by using CSOG in surgery. Comparison of the effect of implant placement after tumor resection with and without utilizing CSOG was studied using FEM. Therefore, models of the mandibular tumor of three human patients were used. After tumor resection continuity defects appeared in the mandible. Four different possible cases (of each patient) of varying gaps between implant and bone were designed and stress & strain distribution were calculated. Hence the effect of improper fitting of the implant to bridge continuity defects of the mandible was obtained. This study proved the advantages of the new technique i.e. CSOG in tumor resection and accurate implant placement in mandibular reconstruction.

2 | MATERIALS AND METHODS

This study was approved by ethical clearance committee GDCH Nagpur (Ref. No. GDCHN/SS/Ethical Commi. Cert./ 96/ 2015). The patients were informed of the study, and their written informed consent was obtained.

Three patients who had been diagnosed with ameloblastoma on the right body of the mandible were studied. Development of the FEA model was based on this actual geometry and the sequential software processing was carried out as follows.

The CT scans of all patient's maxillomandibular region were performed using a spiral CT (Siemens Sensation 16, Munich, Germany). Each patient's scan was comprised of 340 cross-sectional 'cuts' with slice distance of 0.7 mm and image resolution of 512 × 512 pixels. The data of all three patients in DICOM format were imported into the software package Mimics 14.5 (Materialize, Leuven, Belgium) and three 3D diseased CAD models of patient's mandible were generated using segmentation tool and converted into. STL file format (Figure 1).

For all three patients VSP, customized implant, and screws were designed in software 3Matic 8.0 (Materialize, Leuven, Belgium) and exported into. STL file format. Geomagic Studeo 11.0 (Geomagic, North Carolina, USA) design software was used to convert. STL files into IGES file format. Then assembly of mandible, implant, and screws was performed in software Catia V5 (Dassault Systemes).

Four cases (assemblies) of each patient (total 3patient*4cases = 12cases) were prepared for analysis based on the possibilities that may occur at the time of surgery, by resecting the tumor of the mandible at different locations in 3Matic software (Materialize, Leuven, Belgium). Figure 2 shows all four possible cases for the first patient. Likewise, four cases for the each remaining patient were prepared for analysis.

In each patient's 1st case the exact cutting of the tumor using CSOG was prepared. Because of the exact resection of the tumor, the implant was fitted accurately with zero gaps between implant and bone at the junction. All preplanned screws were fixed at planned locations. In each patient's case 2, 3, and 4 resections of the tumor were performed without using CSOG. So overcutting occurred, in case 2 on right side by 3 mm, in case 3 on left side by 3 mm, and in case 4 on both sides by 3 mm. Due to overcutting the implant was not properly fitted at the planned location and a gap was generated in between the implant and the remaining mandibular stumps, in case 2 on right side 3 mm and on left side 0 mm, in case 3 on right side 0 mm and on left side 3 mm, and in case 4 on both side 3 mm (Figure 2). A number of screws on the right side and left side: case 1: 2 and 3, case 2: 2 and 3, case 3: 2 and 2, and case 4: 2 and 2 (Figure 3). Because of the gap in the left side in case 3 and 4 only two screws fitted instead of three screws in all three patients. The main differences between the four cases were gaps between the implant and remaining mandibular stumps, gap locations, and the number of fixing screws. The gap patterns of all four cases (Figure 2) were kept similar for all three patients. Three different implants were prepared for three patients. The same implant was used in all four cases for a particular patient.

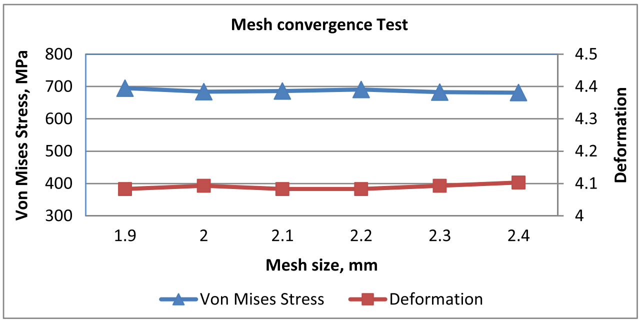

FIGURE 3 Mesh convergence study on von Mises stress and deformation for six mesh size elements

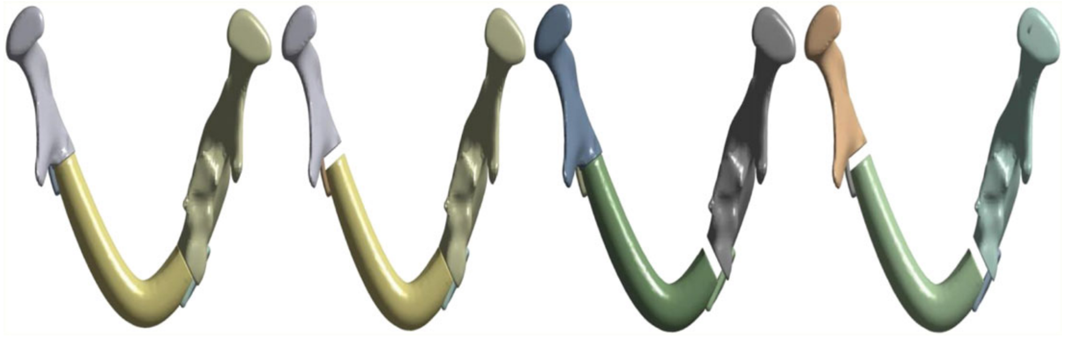

FIGURE 2 The gap between implant and bone on right and left side in case 1: 0 mm on both side, case 2: 3 mm and 0 mm, case 3: 0 mm and 3 mm, and case 4: 3 mm on both sides

3 | FINITE ELEMENT ANALYSES

For FEM analysis of mandibular reconstruction with a customized implant, Ansys Workbench 16.0 (ANSYS Inc. Swanson, Houston, USA) software was used for all 12 cases to evaluate the biomechanical responses. The FE modeling procedure is discussed below.

3.1 | MATERIALS

Material properties of cortical bone, titanium screws and titanium implant were obtained from previously published literature[13] and have been given in Table 1. In all 12 cases, the same material properties for bone, implant, and screws were used (Table 1). In relation to the material properties, the cortical bone, implant, and screws were taken as homogeneous isotropic linearly elastic with different properties. The material selected for the customized implant and screws was titanium due to its high specific strength, biocompatibility, and corrosion resistance.

3.2 | Meshing

The assembly of tumor resected mandible with customized implant model is imported in Ansys Workbench 16.0 (ANSYS Inc. Swanson, Houston, USA) as an IGES file generated from Catia V5 (Dassault Systems) software. In view of the complexity of the mandibular framework, 10-node 3D tetrahedral[14] elements were employed. Before selecting a mesh size for the model, a mesh sensitivity study was carried out to confirm that the employed mesh element size has neither time consuming nor leading to any discretization errors. A convergence study has been conducted on six mesh sizes to obtain an estimate of the variation in the Von Mises stresses and deformations, if any. During meshing, the triangle surface mesher was employed with a program-controlled patch conforming method, i.e., the mesh size was automatically transitioned in the regions where the geometry dimensions have been less than the selected mesh size. Figure 3 shows the results of the mesh convergence study for maximum Von Mises stresses and deformations generated in the case 1 of the first patient, a region of the customized implant for six different element sizes. It can be seen from Figure 3 that the Von Mises stresses and deformations have been similar for the range of the six mesh sizes. In the remaining study, the presented results of all 12 cases were based on the mesh size 2 mm. The total number of nodes and elements in this model used for convergence test were 56739 and 31081, respectively (Figure 4).

3.3 | Loading and contact boundary conditions

In order to define the loading and boundary conditions effectively on a particular location, in all 12 cases, small free-form patches were designed using 3Matic (Materialize, Leuven, Belgium) software as shown in Figure 5. The frictional contact with a coefficient of friction 0.3[11] between the implant's fixation plate & screws, implant fixation plate & bone, implant side & bone was provided. Screws & bone have given a bonded contact. The static structural load conditions were applied for analysis. The loading and boundary conditions were adopted from a previous study (Li *et al.*[13]). The same loading and boundary conditions were used in all 12 cases (Figure 5). The upper parts of the condyles were constrained with fixed support to restrict the displacement in all directions (Figure 5). In all 12 cases 600 N (M. Bakke[15]) force was applied to the left molar, perpendicular to the occlusion surface (Figure 5). The computed parameters of the FEM solutions included Von Mises stress and strain distributions.

TABLE 1 Material properties of different parts of the model[13]

Type of material	Young's modulus (MPa)	Poisson's ratio	Tensile strength (MPa)
Mandible (cortical bone)	8700	0.28	85
Titanium screws	105000	0.3	897
Titanium implant	105000	0.3	897

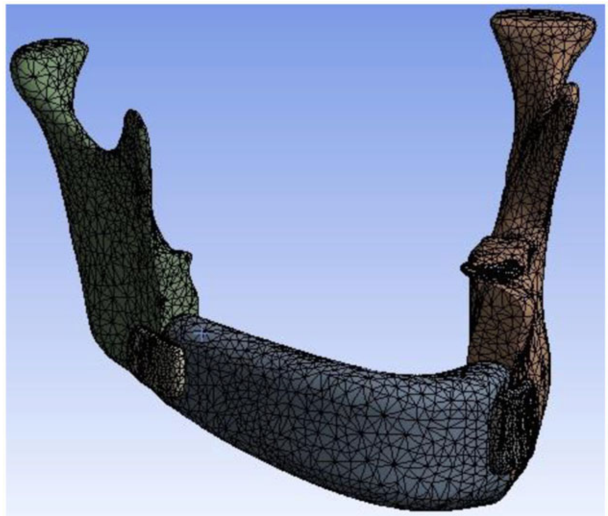

FIGURE 4 Meshing (Nodes = 56739 and Elements = 31081)

3.4 | Statistical analyses

The FEM results were evaluated in all three patient's with using CSOG group (case 1 (zero gaps)) and without using CSOG group (case 4 (maximum gap)) to determine whether the CSOG offered any benefit in the mandibular reconstruction. Paired t-tests were performed using Microsoft Office Excel 2007 to compare the measurements obtained with the two methods (with and without using CSOG). The significance level was set to 0.05.

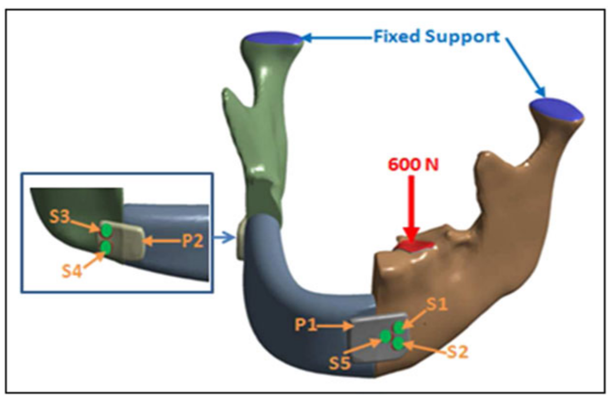

FIGURE 5 Loading and boundary conditions

4 | RESULTS

The simulations has oriented to evaluate the impact of loading (force) on the reconstructed mandible in 12 different cases. These cases contained different gaps between implant and bone. The results have been presented in the form of stress and strain distributions. Higher values considered more critical. Figures 6 and 7 shows the simulation results of stress and strain in the first case.

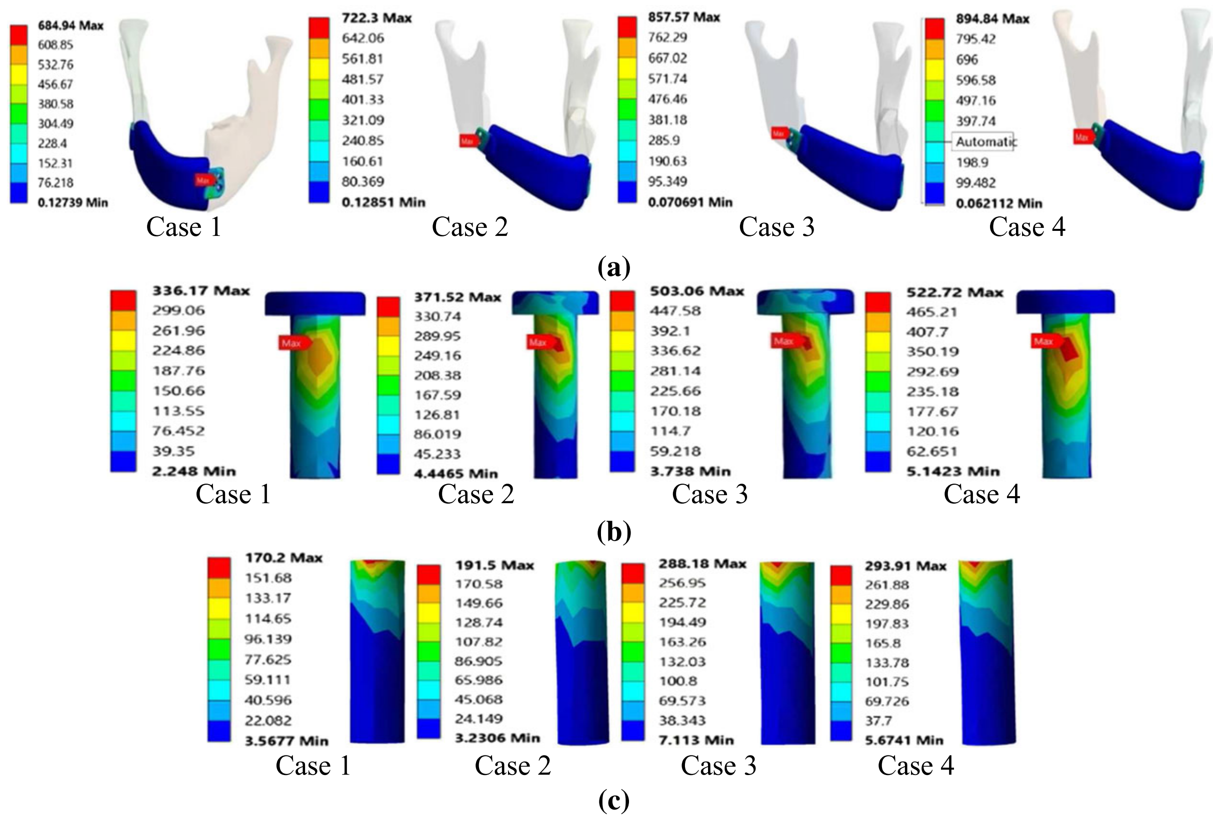

FIGURE 6 Von Mises stress in (a) implant, (b) screw (S3), and (c) screw hole SH3 on the bone in case 1, case 2, case 3, and case 4

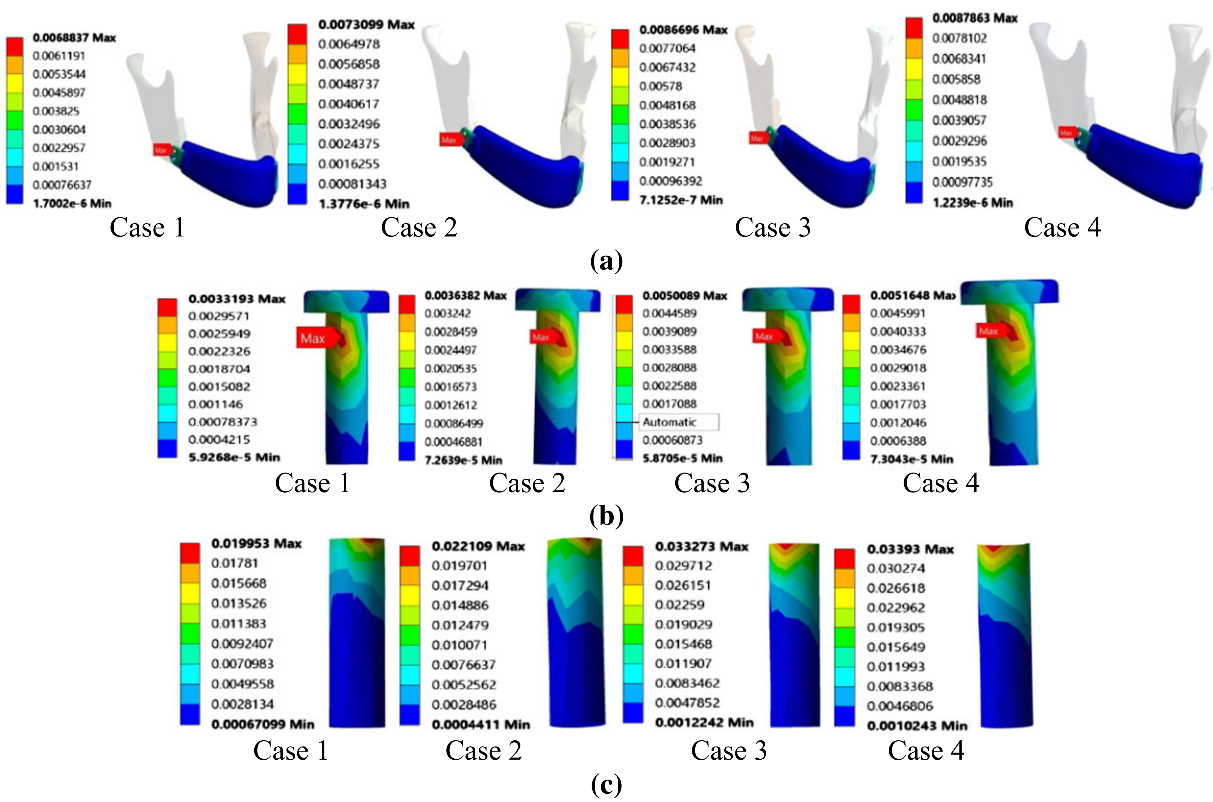

FIGURE 7 Von Mises strain in (a) implant (b) screw (S3) and screw hole SH3 on the bone in case 1, case 2, case 3, and case 4

A comparative summary of simulated results of stress and strain distributions on implants, screws and screw holes on the bone, of all four cases of each patient, have been presented in Figures 8, 9, and 10. Figure 11 shows the mean stress and strain distribution on implants, screws and screw holes on the bone, of all four cases of each patient.

The statistical analysis showed a significant difference in Von Mises stress and Von Mises strain between case 1 (zero gap) (with CSOG group) and case 4 (maximum gap) (without using CSOG group) in mandibular reconstruction.

The Von Mises stress in the implant in case 1 (with CSOG group) was 625.01 ± 58.31 MPa (mean ± SD) which has been significantly lower compared to case 4 (without using CSOG group) 872.46 ± 21.84 MPa (mean ± SD), (P = 0.0072) (Figure 11(a)).

The Von Mises stress in screw S3 in case 1 (with CSOG group) was 312.72 ± 23.30 MPa (mean ± SD) which has been again significantly lower compared with case 4 (without using CSOG group) 544.83 ± 50.52 MPa (mean ± SD), (P = 0.0095) (Figure 11(b)).

Also the Von Mises stress in screw hole SH3 in case 1 (with CSOG group) was 149.35 ± 19.11 MPa (mean ± SD) which has been significantly lower compared with case 4 (without using CSOG group) 283.69 ± 9.69 MPa, (P = 0.0017) (Figure 11(c)).

The Von Mises strain in the implant in case 1(with CSOG group) was 0.0057 ± 0.0012 MPa (mean ± SD) which has been significantly lower compared to case 4 (without using CSOG group) 0.0090 ± 0.0002 MPa (mean ± SD), (P = 0.051) (Figure 11(d)).

Additionally, the Von Mises strain in screw S3 in case 1 (with CSOG group) was 0.00308 ± 0.0003 MPa (mean ± SD) which has been

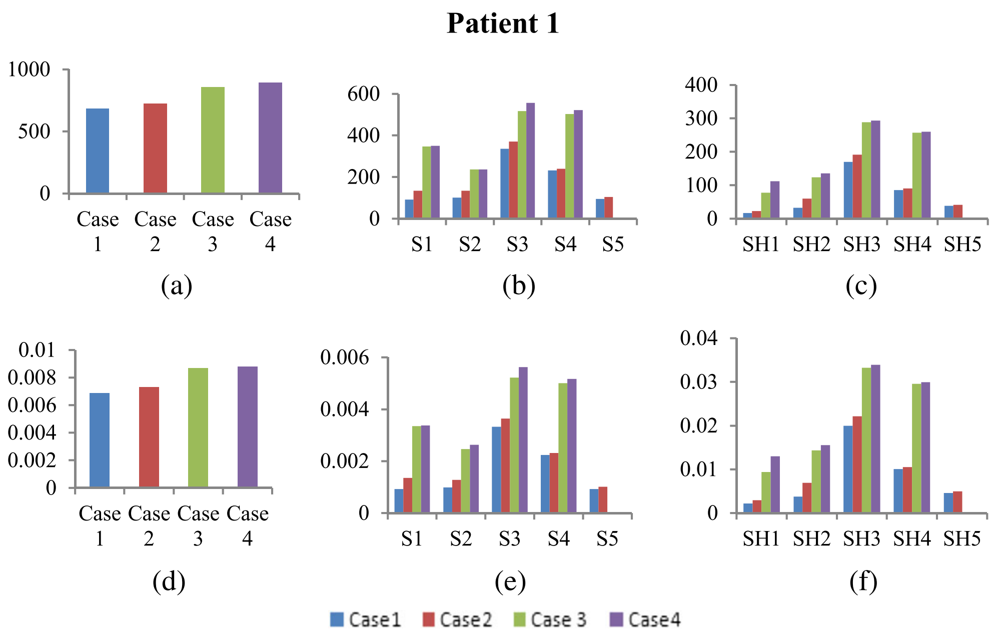

FIGURE 8 Von Mises stresses on (a) implant, (b) screws, and (c) screw holes on bone; von Mises strains on (d) implant, (e) screws, and (e) screw holes on bone

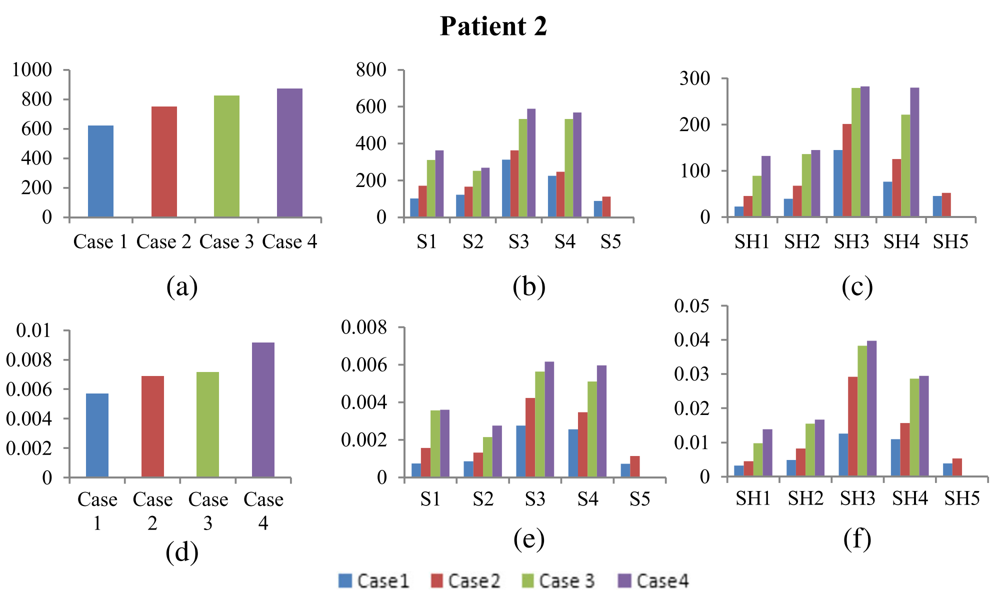

FIGURE 9 Von Mises stresses on (a) implant, (b) screws, and (c) screw holes on bone; von Mises strains on (d) implant, (e) screws, and (e) screw holes on bone

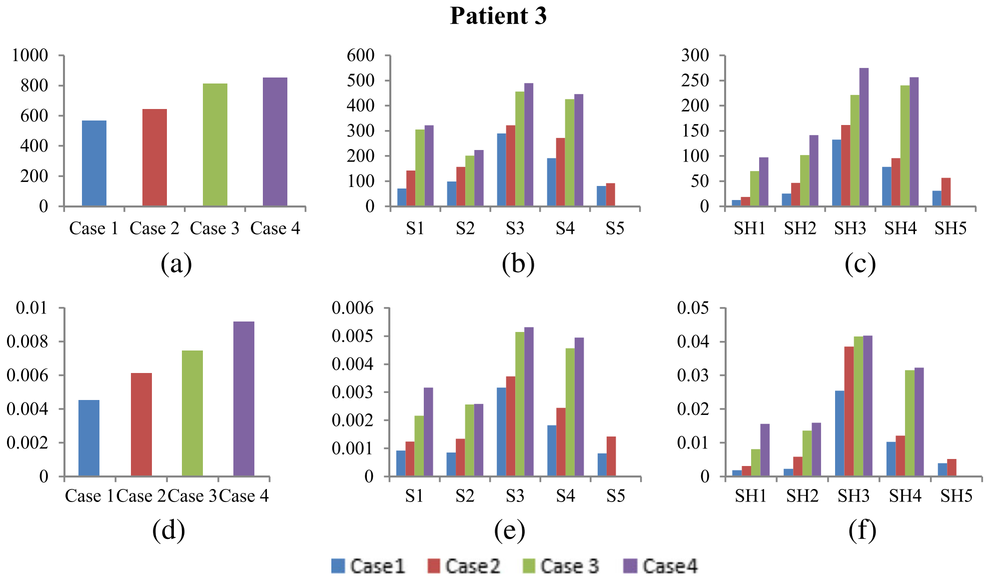

FIGURE 10 Von Mises stresses on (a) implant, (b) screws, and (c) screw holes on bone; von Mises strains on (d) implant, (e) screws, and (e) screw holes on bone

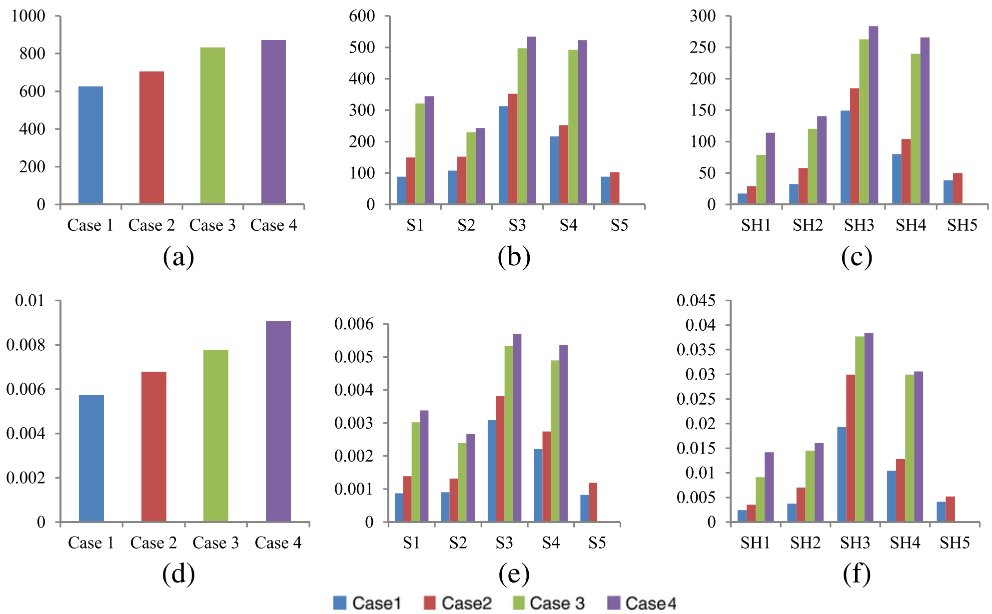

FIGURE 11 Mean Von Mises stresses of all three patients on (a) implant, (b) screws, and (c) screw holes on bone; mean von Mises stresses of all three patients on (d) implant, (e) screws, and (e) screw holes on bone

significantly less than in case 4 (without using CSOG group) 0.00570 ± 0.0004 MPa (mean ± SD), (P = 0.022) (Figure 11(e)).

Furthermore, the Von Mises strain in screw hole SH3 in case 1 (with CSOG group) was 0.01931 ± 0.0065 MPa (mean ± SD) which has been significantly lower compared to case 4 (without using CSOG group) 0.03847 ± 0.0040 (mean ± SD), (P = 0.042) (Figure 11(f)). Tables 2 and 3 has showing the detailed results of statistical analyses.

5 | DISCUSSION

The most common cause for post-operative failure in mandibular surgery is either implant failure, screw failure (fracture due to loads) or the unanchoring of affixing screws. In all patient's case 4, the

TABLE 2 Mean Von Mises stresses

Stresses on	Case 1 (with CSOG group) (mean± SD)	Case 4 (without CSOG group) (mean± SD)	p-value
Implant	625.01±58.31	872.46±21.84	0.007
S3	312.71±23.30	544.83±50.52	0.009
SH3	149.35±19.11	283.69±9.69	0.001

TABLE 3 Mean Von Mises strains

Strain on	Case 1 (with CSOG group) (mean± SD)	Case 4 (without CSOG group) (mean± SD)	p-value
Implant	0.0057±0.0012	0.0090±0.0002	0.051
S3	0.0030±0.0003	0.0057±0.0004	0.022
SH3	0.0193±0.00645	0.0385±0.0040	0.042

mandibular reconstruction with wider gap (in both sides 3 mm i.e. total 6 mm) between implant and bone, induces higher mean maximum stress i.e. 872.46 MPa on the implant at the location of fixation plate P2 which is almost 28% higher when compared with the mean maximum stress 625.01 MPa, developed in case 1 on the implant at the location of fixation plate P1 (Figure 11(a)). Figure 12 showing the comparison of Von Mises stresses and Von Mises strains between case 1 (with CSOG) and case 4 (without using CSOG) on (a) implant, (b) screws, and (c) screw holes on bone in each patient.

In order to achieve a good prosthetic reconstruction, a stable connection between the implant and the mandibular stumps is necessary. This is illustrated as being an essential factor influencing the quality of life.[16] This is attributed to the fact that, in each patient in case 1 because direct contact between implant and bone (i.e. zero gaps) provides increased fixation surface area over the same length when compared with the area of the other three cases i.e. cases 2, 3 and 4 (Figure 13). In Figure 13 the red zone shows the contact area of the implant with bone and the green zone represents the un-contacted area of the implant with bone. Moreover, due to increased fixation surface area in each patient's case 1, all screws fitted at planned locations. Thus, better mechanical fixations of customized implants were possible. The highest mean stress was observed in the screw S3 in case 4 (544.83 MPa) where the gap between implant and bone was maximum. The maximum stresses developed on all the implants and screws of all 12 cases are significantly lower when compared with the ultimate tensile strength (897 MPa) of the titanium alloy. Pure implant or screw strength may not be the critical issue here.

The other parameter to evaluate in mandibular reconstruction of all four cases in each patient is their flexibility, i.e. the capability of each implant to absorb the chewing load. The mandibular reconstruction with a customized implant is considered to be more stable and their interfaces remain intact for longer when implant and screws exactly fit with the bone. This means the mandibular reconstruction using customized implant with minimum gap between implant and bone, induces lower strains in the screws and is considered more flexible.

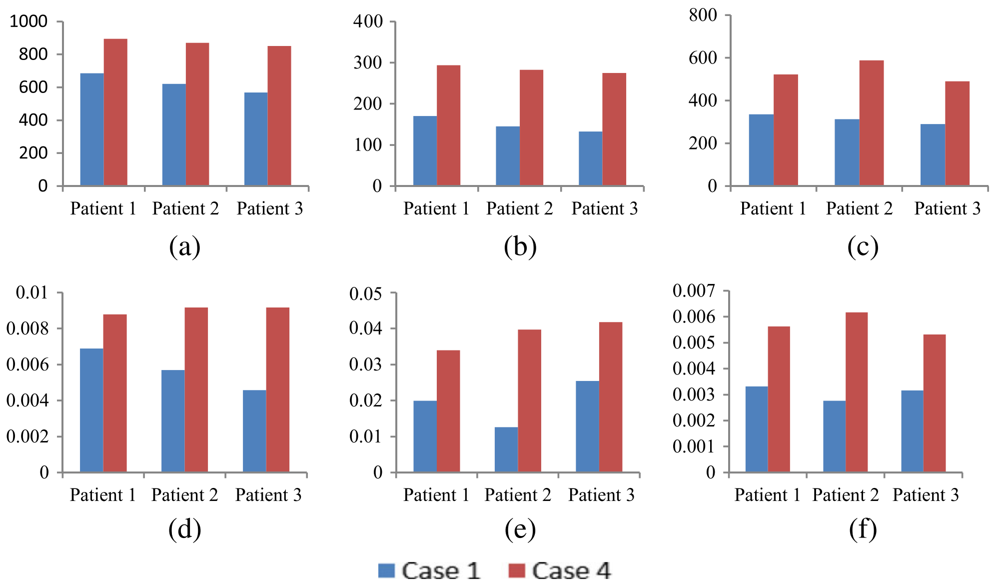

FIGURE 12 Comparison between case 1 (with CSOG) and case 4 (without using CSOG): Von Mises stresses on (a) implant, (b) screws, and (c) screw holes on bone; von Mises strains on (d) implant, (e) screws, and (e) screw holes on bone

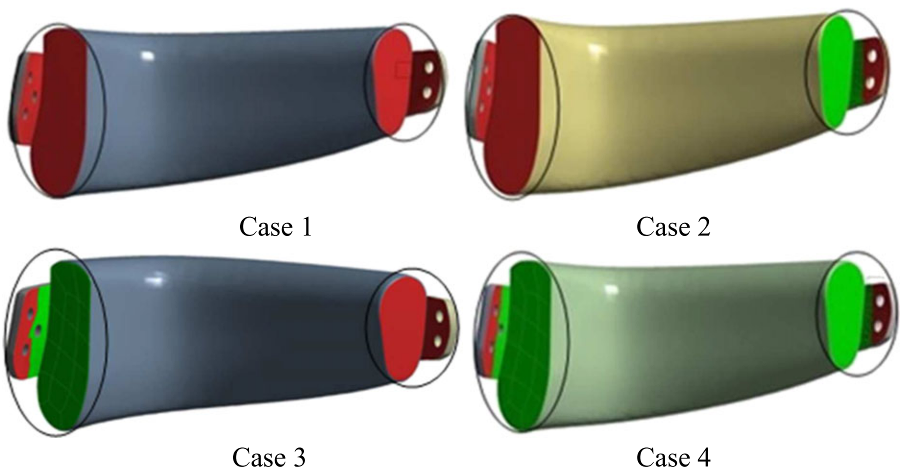

FIGURE 13 Red zone on the implant shows the surface area contacted with the bone. Green zone shows the surface areas which are un-contacted with bone because of overcutting

As seen in Figure 11(d), the maximum mean strain developed inside the screw hole on the implant in case 4 is (0.0090), case 3 (0.00778) and case 2 (0.00677) when compared with case 1 (0.0057). This implies that relatively less load will be transferred; resulting in lower strains developed in the screws in the 1st case, thus providing better flexibility. These results indicate that the chances of loosening the screws are lower in the 1st case due to lower mean maximum strain developed when compared to 2nd, 3rd, and 4th case.

The difference among gaps between implant and bone becomes apparent when analyzing for strains induced in the bone in the screw holes. Our results indicate that case 1, where the gap is zero between implant and bone, induces lower mean maximum strain in the bone at the anchoring sites (0.01931) when compared with case 2 (where the gap on the left side is 0 mm and right side is 3 mm) (0.02993), case 3 (where gap on the left side is 3 mm and the right side is 0 mm) (0.03770), and case 4 (where gap on both sides is 3 mm) (0.03847) (Figure 11(f)). Thus, it seems prudent to allow for a certain amount of flexibility in the implant for them to be able to better absorb cyclic stresses, and thereby minimizes the load transferred to the affixing screws. This would ensure that the implant's plate–screw and screw–bone interface remains as stable as possible, especially in the initial healing stage.

It is, therefore, not only sufficient to merely cater for the strength of the implant and screws, but also is vital during precise tumor resection and accurate implant fixation on planned location, the number of screws, their orientation, and arrangement of screws fixation.

6 | CONCLUSIONS

In this work, a relatively new approach to mandibular reconstruction with customized implants is implemented. A comparative study of 12 cases (3 patients * 4 cases = 12 cases) having different gaps between implant and bone was analyzed using FEM. In the first case, tumor resection was performed using CSOG giving zero gaps between the implant and remaining stumps on both sides. In the second, third and fourth case, tumor resection was performed without using CSOG. Hence overcutting occurred on left and right side in case 2: 0 mm and 3 mm, case 3: 3 mm and 0 mm, and case 4: 3 mm on both sides. Using FEM biomechanical behavior of four cases of mandibular reconstruction (having total gap 0 mm, 3 mm, 3 mm and 6 mm between implant and bone) for each patient was examined on the basis of stability through computation of stress and strain distribution. The most common cause for post-operative failure in mandible reconstruction is either implants fixation plate failure (related to stress) or loosening of the screws (related to strains). The mean maximum stress induced in the mandibular reconstruction of the first case (625.01 MPa) has been significantly less when compared with the second, third, and fourth case (705.46 MPa, 831.17 MPa and 872.46 MPa). However, the maximum stresses on the implants and screws of all four cases of each patient are well below the failure limits of titanium alloy. This indicates that pure implant and screw strength is not the critical issue here. Furthermore, it is observed that the maximum stresses developed alone have not sufficient to evaluate the effect of gaps between the implant and mandibular stumps. The load transferring capability (flexibility) needs to be accessed as well by observing the strain developed in screws in all four cases of each patient. The FEM results reveal that the maximum gap between implant and bone have been more susceptible to loosening of the screws due to higher mean strains concentrated on the screw hole (37% higher) under same loading conditions when compared with zero gaps (first case).

The analysis results indicate that the gap between implant and bone due to overcutting has more sensitive in mandibular reconstruction. If the gap goes on increasing it results in increasing stresses and strains in mandibular reconstruction. These FEM results enhance the understanding of mandibular reconstruction when considering the resection of tumor and implant placement. It can be suggested that zero gaps between implant and bone after tumor resection can significantly improve the stability and safety of the mandibular reconstruction. In future, we aim to evaluate the designs further under dynamic conditions incorporating a more detailed model.

In conclusion, when placement of a customized implant in mandibular reconstruction, it is paramount that the focus should be on accurate tumor resection so that zero gaps between the implant and resected mandibular stumps are achieved, while at the same time all screws should be fixed at planned location on the bone.

CSOG helps to minimize the different errors in tumor resection and implant fixation. Also customized surgical guide (CSG) helps to reduce the surgery time and improve the surgeon's confidence.

This technique of CSGs can be used in different surgeries for accurate resection of bone or accurate drilling in the bone or accurate positioning of the resected bone. In each case for a particular purpose a related CSGs has to be developed followed by VSP. Generally these types of CSGs can be used in different surgeries like total hip replacement, total knee replacement, dental surgery, craniomaxillofacial surgery, corrective osteotomy, orbital implant placement, orthopedic oncology, bone resection and allograft reconstruction, deep brain stimulation, mandibular distraction osteogenesis…etc.[17]

The main limitation of this CSG technique is the cost. A CSG requires costly techniques for its development like radiology, image processing, VSP, CAD, RP technique, postprocessing and sterilization) to fabricate RP-assisted CSGs, so that it can be used in routine clinical practices.

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