

Stress in the Mandible with Splinted Dental Implants caused by Limited Flexure on Mouth Opening: An *in vitro* Study

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ABSTRACT

Aim: The aim of this study was to evaluate the stress developed in the bar connecting implants and in the mandible as a result of the elastic deformation of the mandible during mouth opening when using a finite element method (FEM).

Materials and methods: A three-dimensional model of an edentulous mandible was generated based on the computer tomography (CT) data of a patient. Two cylindrical implants (diameter 4.3 mm, length 13 mm) were inserted in the area of the mandibular canine, premolar and molar in the mandibular model. Implants were connected with a rigid bar (width 2 mm, height 3 mm), and mouth opening was simulated on the three-dimensional (3D) model. The location and magnitude of maximum von Mises stress that occurred in the mandible and in the bar were estimated.

Results: The highest stress level in the mandible (4.5 GPa) and in the splint (32 GPa) was measured in the longest fixed partial denture with the implants in the mandibular left canine and left second molar position. The maximum stress in the bone was measured distal to the splinted implants.

Conclusion: Since, great distance between splinted implants caused high stress during mouth opening, due to mandibular deformation, the use of a short span fixed partial denture supported by implants in the molar region of the edentulous mandible is probably more advantageous.

Keywords: Finite element analysis, Mandibular distortion, Splinted implants, Stress.

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INTRODUCTION

The success rate of implants depends on good patient selection, implant-prosthetic planning, surgical methods and types of prosthesis, optimal occlusion, articulation, load distribution and proper maintenance. Whether implants should be splinted to enhance their load bearing capacity functioning as abutments of prostheses is still an unanswered question in prosthetic dentistry. By using dental implants in the premolar and molar region patients can avoid wearing a removable partial denture thus, ensuring greater comfort. In an optimal situation the bone morphology or augmentation methods allow placement of implants for all

of the missing teeth. However, this is not always possible due to anatomical or financial restrictions. For these patients fixed partial dentures are provided. The length of the fixed partial denture depends on the number of missing teeth and the length of the edentulous ridge.

A flexure of the mandible during the movements was examined in several studies. Gates and Nicholls¹ measured a 0 to 0.3 mm decrease in the mandibular arch width between the lower first molars during opening and 0.1 to 0.5 mm at maximum protrusion. In edentulous subjects a relative movement of up to 420 μ m, between osseointegrated implants replacing the premolars during active opening and a medial convergence of the condyles during opening between 0.0 mm and 1.5 mm, was reported.²

The flexure of the mandible may have a greater impact on implant therapy since osseointegrated implants have no periodontium. They are unable to compensate for the flexure of the mandible during movements. The stress in the fixed prosthesis may cause screw loosening or break; bone resorption in the surrounding bone may occur ultimately resulting in implant loss.³ A splint connecting the implants is more rigid than the bone; therefore, high stress may be generated during mandibular movements. An *in vitro* investigation by Hobkirk and Havthoulas⁴ stated that the loads, when six implants were connected, were more widely spread and greater extrusion forces occurred. It was concluded that the mandibular deformation during function is an important factor in the case of implant-retained fixed prosthesis. In a three-dimensional finite element model, Zarone et al⁵ examined the effect of fixed partial denture designs of differing lengths on the stress distribution in the superstructure during mandibular movements. A greater amount of stress in the more distal implants was recorded. Periimplant marginal bone resorption may be the consequence of unfavorable mechanical factors, e.g. load transmitted by the implants to the surrounding bone, or stress concentration in the periimplant region.⁶

Splinting implants is unavoidable in some situations, namely when anchoring removable partial dentures by bar retention or by providing a fixed partial denture for replacement of missing teeth. However, the effect of splinting on the different structures involved in such a system (fixed partial denture, implant and abutment),

regarding the bone deformation during the masticatory function is not yet entirely clear. Finite element analysis proved to be an appropriate method to simulate the biomechanical factors in implant prosthetics.⁷

The aim of the present study was to evaluate the stress levels that occurred in the mandible and in the rigid bar connecting the implants inserted in the mandibular canine, premolar and molar region during mouth opening as a result of the elastic deformation of the mandible without any load on the implants or on the bar using finite element method (FEM).

MATERIALS AND METHODS

A three-dimensional model of an edentulous mandible was created by a surgical planning system⁸ (TraumArt Ltd, Hungary). The system was originally developed for trauma patients to provide a computerized planning tool for the fixation of multiple fractures, particularly in patients where complex bone geometry and high loading and material stress are involved. This software tool is able to construct a 3D bone model from the patient's computer tomography (CT, GE Lightspeed, General Electric, USA) images and offers a library of 3D implants which can be inserted for virtual fixation. The tool calculates the biomechanical behavior of the surgical plan by the finite element method which provides information of the deformations and stress levels of the bone and implants before the actual surgery takes place. The system was used successfully in several real-life situations for planning surgical interventions.

In the present study, the CT data of a 64-year-old female patient was used which contained 214 slices of her facial region. The distance between slices and the slice thickness were both 0.625 mm, the pixel spacing was 0.33 mm in both x and y directions. To create the three-dimensional model the following steps were performed. First, the mandible was segmented, which is basically a conversion into a binary image where only the background and the bone is represented with zeros and ones respectively. The segmentation algorithm was based on fuzzy connectivity.⁹ This allowed a fast 3D semi-automatic segmentation since, only a few seed-points were required to achieve a reasonably good segmentation. This method sometimes failed to find the inner structure of the bones due to the fact that cancellous bone tissue is much less visible on the CT than cortical tissue. This resulted in holes and cavities in the segmentation which were corrected by hole-filling and isolated voxel removal. Both methods are automatic and do not require user interaction.

Next the surface of the segmented voxels was determined with the Marching Cubes algorithm¹⁰ which

converts the volumetric data to a surface triangle mesh. The generated triangle mesh of the mandible contained around 263000 faces and 131000 vertices. This mesh was further simplified and smoothed¹¹ to increase later processing performance. The finite element mesh was directly generated from the geometry obtained after smoothing; therefore, the size of the mesh had a major impact on the runtime of the analysis. After simplification the model contained only 9000 faces and 4500 vertices. The steps described in Figure 1. The study was approved by the Ethics Committee of the University of Szeged.

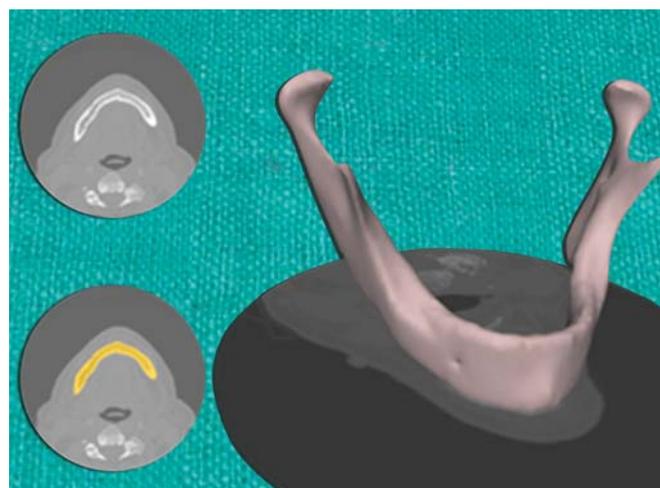


Fig. 1: Creation of the mandibular model. A single slice of the mandible CT scan (top left), the same slice after segmentation (bottom left), the 3D surface mesh (right)

Two cylindrical Camlog[®] (Camlog Biotechnologies AG, Basel, Switzerland) implants were inserted into the model mandible. The measurements were as follows: The height of the mandible 24.6 mm, the width in the crestal area 6.5 mm. The diameter of the implants was 4.3 mm and the length was 13 mm. Positions of the implants in the left side of the mandible were: Canine-second premolar, canine-first molar, canine-second molar, and second premolar-first molar, second premolar-second molar area. The locations of the implants were determined by placing the model of a similar but dentate mandible over the edentulous one. Interimplant distances were 11.5, 18.3, 23.5, 14 and 19 mm respectively, which were measured between the centers of the implants. There was no relative movement between the implant and the bone, the fixation was entirely rigid representing the interface of osseointegrated implants. The implants were connected by a bar 0.5 mm above the cortical bone. The splint had a rectangular shape in cross-section with the dimensions of the width 2 mm, height 3 mm. The implant and the abutment were regarded as 1 unit; the connection with the bar was created without any gap. The material of the bar was set to be identical to the implants.

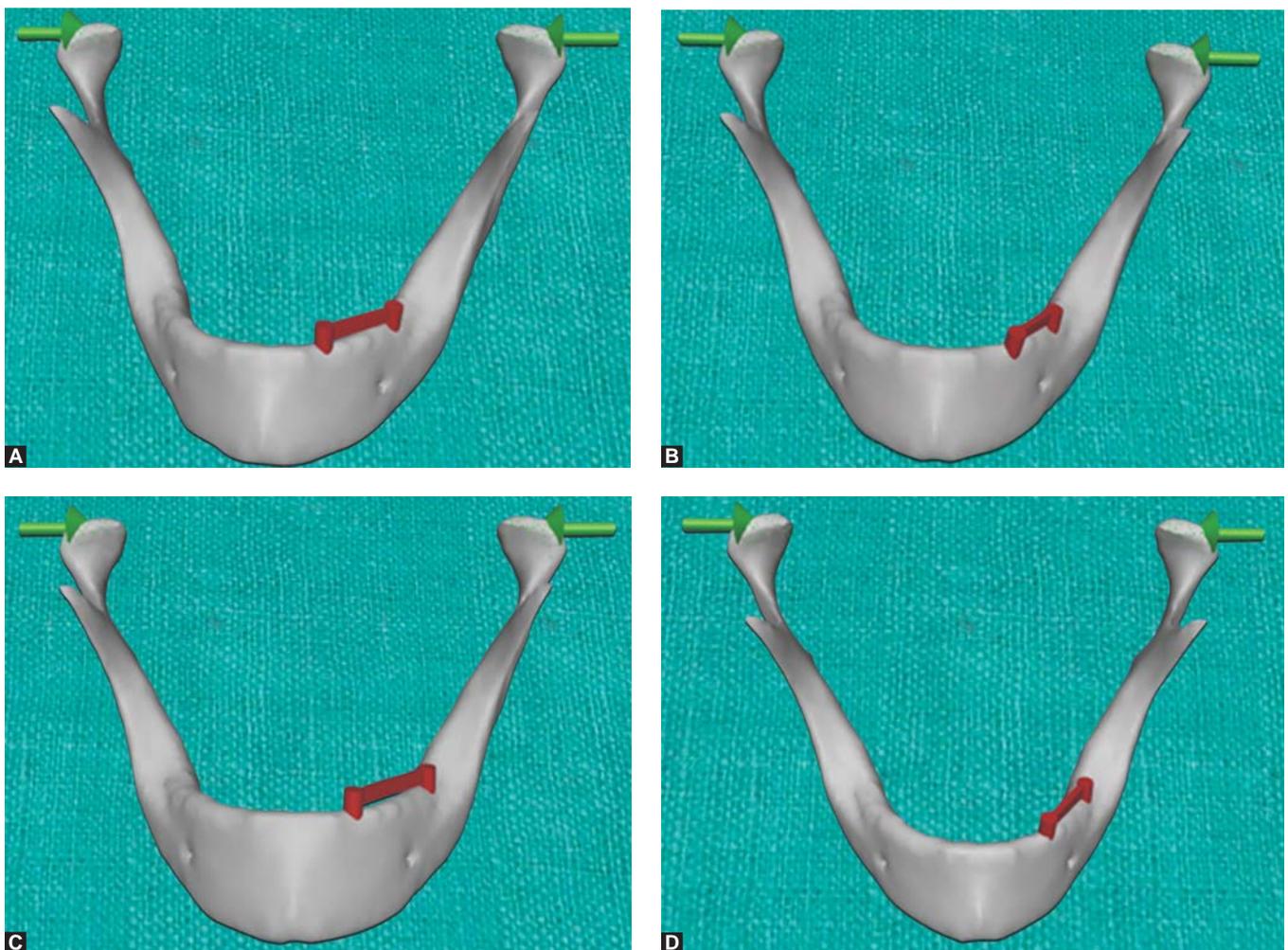
After creating the model and ‘inserting’ the implants the mandible opening simulation was carried out. On opening, the condyles moved toward the sagittal plane 1mm on each side. This meant a decrease in the mandible arch of 0.5 mm in the region of the first molars. This medial convergence range was based on data from the literature.¹² Methods applied in the studies and the results varied a little; a presumed 0.5 mm convergence between the first mandibular molars at maximum opening was used.

For the data analysis, a model was made using material properties taken from the literature.¹² The cortical bone was defined by elastic modulus (E) = 11 GPa and Poisson ratio (ν) = 0.28, for the implants E = 200 GPa and ν = 0.33 were used respectively. The Poisson’s ratio of commercially pure titanium was 0.33 and Young’s modulus was 105 GPa.¹³ The finite element mesh was created from shell elements. Shell elements are 3-node triangular elements used for the analysis of three-dimensional structural models. Six degrees of freedom per node are allowed for structural analysis (3 translational and 3 rotational components). The elements are assumed to be isotropic with constant thickness.

One millimeter thickness was used for the geometry of the mandible, which is similar to the cortical bone tissue.

The finite element mesh of the mandible was generated from the geometry obtained after simplification. For every triangle used, a 3-node shell element was inserted into the finite element mesh. The geometry of the implants was available as CAD (computer-aided design) models, so the generation of finite element meshes could be undertaken using the CAD system. Additional 2-node elements were used to simulate the connection between implants and the mandible. The software used for the analysis was the Cosmos/M system (originally by Structural Research and Analysis Corporation, now Solid Works). For both condyles a 1 mm displacement was defined in the direction of the sagittal plane. The constrained points are shown in green in (Figs 2A to D). The green arrow indicates the direction of the displacement.

Using the above-described setup the following simulations were performed. First, the intensity of the maximum von Mises stress in the unsplinted mandible was measured during opening. Then the maximum stress with splinted



Figs 2A to D: Splinted implants in the left side of the mandible. The positions of the implants from left to right: Canine—first molar, canine—second molar, first premolar—first molar, first premolar—second molar (A: 33-36, B: 33-37, C: 34-36, D: 34-37)

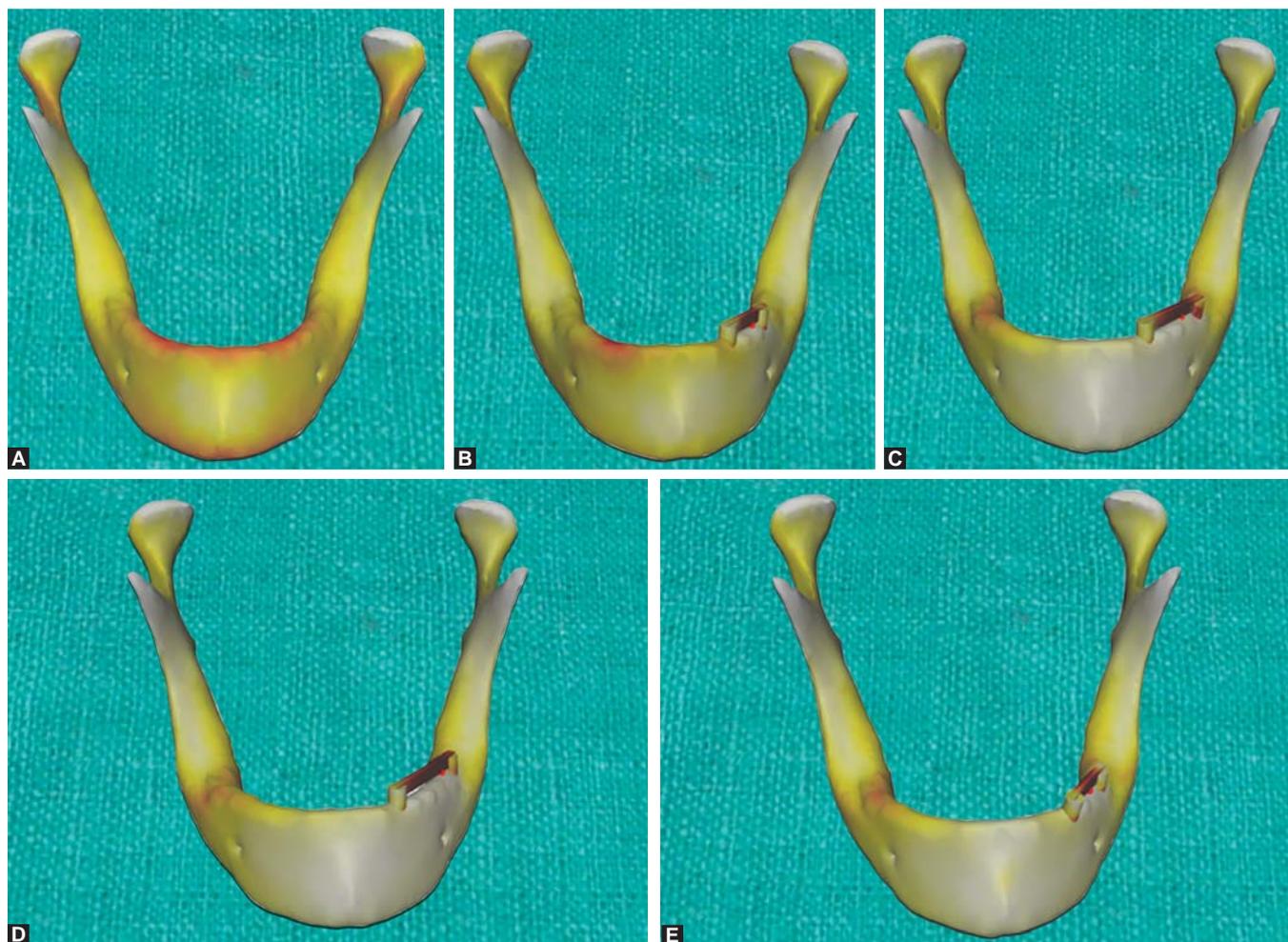
implants in the mandible occurring in the bar and in the reconstructed mandible was measured (Figs 3A to E).

RESULTS

On opening, the lingual displacement of the mandible model measured at the position of the first molar was smaller (0.43-0.47 mm) when implants were splinted (Table 1) vs unsplinted implants (0.5 mm). The more distal the position of the second implant the smaller the displacement

(0.43 mm) of the mandible model; the correlation was almost linear.

After simulating the opening movement of the edentulous mandible model there was some measurable stress level in the cortical bone (1.4 GPa) in the area at the base of the symphysis. This was taken as a baseline (100%), and the stress generated in the mandible with splinted implants were compared to this level. During mouth opening the maximum stress developed in the cortical bone was 40 to 220% bigger with splinted implants (Table 1). The highest



Figs 3A to E: Visualization of the mandible deformation without (A) and with bars of different lengths (B,C,D,E). Colors represent stress values on the surface (dark colors show the bigger stress values)

Table 1: Displacement of the simulated mandible at the tooth 36 and maximum stress levels developed in the splint and in the mandible at mouth opening

Implant position in the mandible left side	Distance between implants (mm)	Displacement of the mandible at the tooth 36 (mm)	Max. stress in the splint (GPa)	Max. stress in the mandible (GPa)
Without implant		0.50		1.4 (100%)
Canine-2nd premolar	11.5	0.47	20.0	2.0 (+ 40%)
Canine-1st molar	18.3	0.45	30.0	3.0 (+120%)
Canine-2nd molar	23.5	0.43	32.0	4.5 (+220%)
1st premolar-1st molar	14.0	0.45	30.7	2.8 (+100%)
1st premolar-2nd molar	19.0	0.44	30.8	3.5 (+150%)

stress value (4.5 GPa) in the mandible model was measured with the longest fixed partial denture when the implants were in the canine and second molar position. The maximum stress in the bone was presented distal to the splinted implants. A stress level as high as 8 to 11 GPa could be observed on the contralateral side of the mandible as well. This stress occurred in the area of the right canine and premolars when the bar connected the implants in the position of the left canine and second premolar in the mandible and more distally in the case of the other splints.

The maximum stress was located at the distal third of the bar for all implant situations and varied between 20 and 32 GPa. The maximum stress level in the bar was measured when the implants had the greatest distance between them. The stress level increased in the longer bars in a greater degree when the mesial implant was in the position of the canine (33), and less when the mesial implant was in the position of the first premolar (34). No stress was detected in the mandible under the bar between the implants.

DISCUSSION

The described method has its limitations, this is mainly due to approximation steps performed to create the 3D model and the limitations introduced by the finite element method. The errors in the 3D geometry originate from the CT image, the segmentation and the mesh generation. Using elastic material model with isotropic material properties during the finite element analysis is also an approximation. Discarding the cancellous bone from the FE calculations is a valid approximation since cortical bone is 10 times stiffer than cancellous bone.

Surgical planning software was used to create the 3D model and the finite element mesh from CT images of an edentulous patient. The software was also designed to allow the insertion of various implants into the surgical plan. Two implants at different distances were connected by a splint, representing a fixed partial denture where the connection of the superstructure and the osseointegrated implants was rigid. There was no possibility for the relative displacement between the splinted implants; therefore stress was generated in the model when mouth opening was simulated. The more distal the second implant was positioned, the less mandibular flexure allowed by the bar. The fixed partial denture connecting the implants limited the flexure of the mandible and the relative displacement of the implants. This phenomenon appeared as a stress in the connecting bar and in the mandible. The stress that occurred at opening may have the result in bone resorption around the implant and may eventually lead to loss of the implant.³

In the present study it was seen that the simulation of mouth opening caused a stress in the bone and also in the fixed partial denture. The stress was larger in the case of longer span fixed partial dentures and when the first abutment was situated more mesially. The reason for the stress is that the implant material and the connecting bar are more rigid than the bone; the splinted implants reduced mandibular deformation, which resulted in higher stress in the bone tissue distal to the splinted implants.

Short span fixed partial dentures, as in the present model, allowed more physiologic bone flexibility during opening, and caused lower stress in the superstructure than long rigid connection between the implants.⁵ The optimal distance between two splinted implants was found to be 11 mm in a finite element model when examining distances of 5, 10 and 20 mm at loading in different directions and quantities¹⁴ while the elasticity of the mandible was not taken into consideration. Although through a different model analysis, a similar optimal distance of two implants was found in the present study; stress measured in the cortical bone or in the bar was the smallest in the case of a distance of 11.5 mm when measured between the centers of the implants.

There was no stress detected in the mandible under the bar, which can be explained by the fact that the bar being stiffer than the bone relieves it from the stress that occurs due to the mandibular deformation.

Within the limitation of this *in vitro* study, it could be shown that mandibular elastic deformation in case of splinted implants in a finite element model had an effect on the stress levels occurred in the mandible and in the bar connecting the implants. Longer distance between splinted implants in the mandible resulted in higher stress in the bone than at a shorter distance. The FE method with simulated mandibular flexure showed that the use of short span fixed partial denture in the molar region of the edentulous mandible decreased stress levels.

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