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NUMERICAL ANALYSIS OF GEOMETRIC CHARACTERISTICS OF MANDIBLE FIXATION PLATES MADE OF HYDROXYAPATITE STRUCTURES

NUMERIČKA ANALIZA GEOMETRIJSKIH KARAKTERISTIKA FIKSACIONIH PLOČICA DONJE VILICE IZRAĐENIH OD HIDROKSIAPATITNIH STRUKTURA

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- fixation plates
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- numerical analysis
- mandible

Ključne reči
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- metoda konačnih elemenata
- fiksacione pločice
- polilaktid (PLLA)
- numerička analiza
- mandibula

Abstract

Bone defect reparation can be successfully performed with biomaterials based on synthetic hydroxyapatite (HA). In addition to collagen, hydroxyapatite is the basic structure of bone tissue and it is expected to have excellent biocompatibility. This study presents analysis of HA properties, its biocompatibility, and its enhancement due to the reinforcement by biocompatible polymers. This implies the use of bioresorbable poly-L-lactide (PLLA) and bioresorbable HA to produce a unique high-porosity composite HA/PLLA with mechanical properties very similar to bones.

Using the finite element method, the influence of the mandible fixation plate geometry on the stress distribution is analysed. There are several existing fixation plate models but also a few new-design plates. The mandible model is made with several assumptions. The total effect of occlusal forces is approximated by a 150 N load in mandible muscle.

Analysis shows that the best characteristics appear in large surface fixation plates, but the minimum surface plate design with two rows of screws is selected due to the connecting problem in the bone-implant system (naturally uneven).

Izvod

Reparacija defekata kostiju može se veoma uspešno obaviti biomaterijalima na bazi sintetskog hidroksiapatita (HAP). Hidroksiapatit je pored kolagena osnovna struktura koštanog tkiva te se sa razlogom mogla očekivati odlična biokompatibilnost. Predstavljena je analiza osobina HAP, njegova biokompatibilnost kao i unapređenje osobina zahvaljujući ojačavanju biokompatibilnim polimerima. To podrazumeva upotrebu bioresorbilnog poli-l-laktida (PLLA) i bioresorbilnog HAP kako bi se dobio jedinstveni visokoporozni kompozit HAP/PLLA sa mehaničkim osobinama veoma sličnim kostima.

Primenom metode konačnih elemenata, analizirane su naponske karakteristike različitih oblika fiksacionih pločica donje vilice. U pitanju je nekoliko postojećih modela ali i predlozi novog dizajna. Model vilice napravljen je uvođenjem nekoliko aproksimacija. Dejstvo okluzionih sila aproksimirano je opterećenjem viličnog mišića od 150 N.

Analiza pokazuje da najbolje karakteristike imaju fiksacione pločice velikih površina, ali je zbog problema povezivanja u sistemu kost-implant (prirodno neravno), odabran dizajn pločica minimalne površine sa dva reda zavrtnjeva.

INTRODUCTION

Internal fixation using plates and screws is the best way for fracture fixation. Materials used for this purpose in most cases are Ti alloys, Co-Cr alloys and stainless steel. The problem of rigid internal fixation occurs due to the large difference between bone and implant modulus of elasticity, which results in pressure on bone tissue that can cause bone atrophy and loss of bone mass, primarily below the implant and sometimes even more widely. The new approach is based on biodegradable implants, but the problem occurs because the rate of degradation does not correspond to the rate of regeneration of bone fracture, and also mechanical properties do not meet the desired criteria.

C. Van et al. analysed the effect of fixation plates and intramedullary systems on vascularity in part of the bone where healing occurs. Torsion tests performed by two fixation plates tests showed more torsional resistance than intramedular fixation methods, but this difference had disappeared after 120 days. Fixation plates lead to better mechanical characteristics in the early healing stage. The time required for healing is the same for both methods, /1/.

B. Qiao et al. studied characteristics of fixation plates composed of ternary nano-hydroxyapatite/polyamide 66/ glass fiber composite. Mechanical properties and biocompatibility were tested under in vitro conditions. The results confirmed that adhesion meets the criteria. Fixation of the fractures with this type of composite plates shows less rigidity and satisfactory loading ability compared to fixation using titanium plates. Research has shown that the matrix of hydroxyapatite composite is suitable for osteogenesis and cells differentiation, /2/.

Figure 1 shows a part of the surgical procedure of the mandible fracture (Fig. 1a) and the installed plate (Fig. 1b). The material of fixation plate is titanium, /3/.

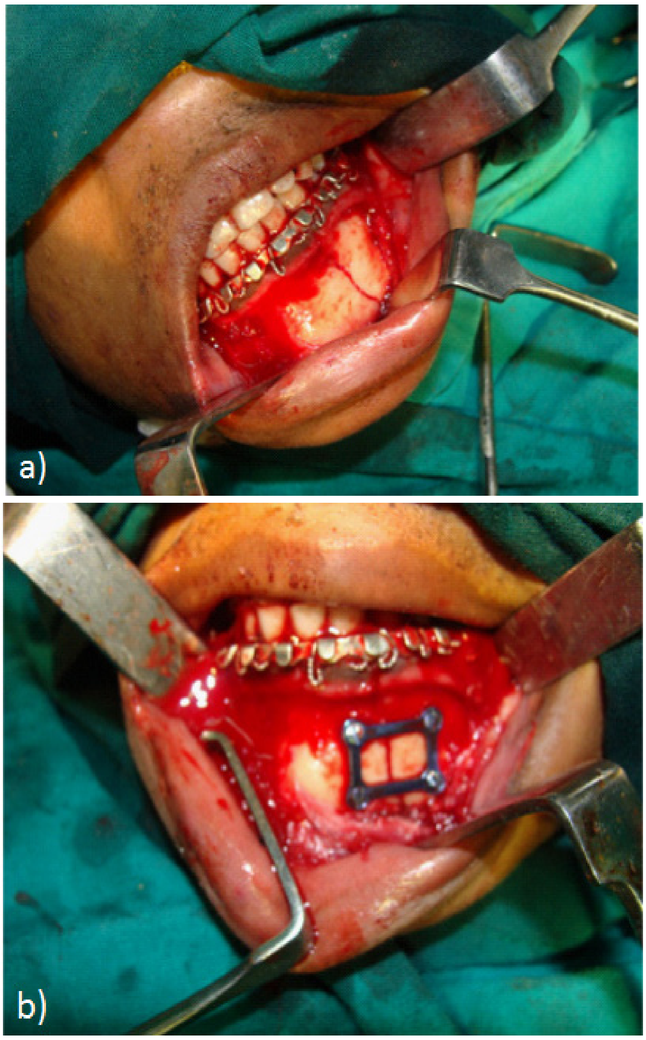

Figure 1. Installation of fixation plate, /3/.

Tested nano-hydroxyapatite/polyamide composite shows yield strength of 184 MPa and modulus of elasticity in the range of 10-20 GPa. In comparison with titanium plates, the bending and torsional resistance of nano-composite fixation plate was reduced by 34.1 % and 56.8 %, respectively. The lower stiffness of the nano-composite fixation reduces the pressure on the bone, allowing the bones to carry more load while creating a new bone tissue, /2/.

F. Atik et al. studied various forms of titanium fixation plates in order to determine which shape provides the highest quality of fixation and load capability. The tested plate shapes were double straight plate, square plate, and Y plate. The highest stress values were detected on Y plate, while the lowest stress values were detected on the double straight plate, which achieved better fixation and load capability, /4/.

Thanks to the applied numerical analysis, it is possible to examine the geometry of the observed material in terms of optimization before the implant is made. This approach leads to time savings, materials and money.

Porous biomaterials have the potential for extensive use in reparation and replacement of bone tissue. The porous structure allows the cells to penetrate into material and allow good vascularity of the new tissue. To be able to predict material behaviour, it is necessary to pay attention at the elementary level. I. Balać et al. generated a two-phase and a three-phase model of unit volume. The characteristics of the two-phase model correspond to predicted characteristics obtained by calculations based on theoretical equations. The importance of this model is that it can be suitable for designing porous materials with the desired characteristics in terms of shape and size of pores, /5/.

HYDROXYAPATITE STRUCTURES AND HA/PLLA COMPOSITE

Bioceramics are used in a wide range of medical systems. According to the type of interaction with the surrounding tissue, bioceramics can be classified into bioactive, bioresorbable, and bioinert.

The most frequently used bioactive ceramics are hydroxyapatite and bioactive glass. It has not yet been fully explained why bioactive ceramics connect directly to the bone. The most probable reason is that apatite is the dominant inorganic phase of hard tissues. The cortical bone is a composite made of collagen and apatite. Collagen provides good stretch resistance and the apatite structure provides resistance to pressure.

Bone tissue consists of calcium phosphate (69%), collagen (20%), water (9%) and other elements (2%). Hydroxyapatite is widely researched as material for reparation of damaged tissues. The term hydroxyapatite usually refers to $Ca_{10}(PO_4)_6(OH)_2$, although the term 'apatite' originally means a much broader class. The general formula for apatite is given by: $M_{10}(RO_4)_6X_2$, M = Ca, Na, K, Mg, Sr; RO_4 = PO_4, HPO_4, CO_3 and X = OH, Cl, F, CO_3, /6/.

There are three types of ceramics used to reconstruct bone defects - dense, porous and granular. Dense ceramics are used in places where high mechanical pressure is required. The construction of dense ceramics provides physical strength, while porous ceramics have a property to provide flexibility. Granular ceramics are most often used in the form of fillings, /7/.

Porous HAP ceramics are primarily used as a replacement for damaged bone tissue, because the pores allow connecting of the contact surfaces. It is obvious that reducing the density of the implant reduces mechanical properties and this type of material cannot be used when implants with great mechanical properties are required. The sintering process will cause a decrease in pore size which will directly alter mechanical properties, /7/.

Polylactide (PLA) is characterized by being completely non-toxic, and this has been confirmed by studies that have observed products of degradation. Mechanical properties depend on the material structure and can vary in a wider range (modulus of elasticity 0.5-10 GPa). PLLA with high porosity (93%) in the form of foam with pore size of about 500 μm has found wide use in tissue engineering as a polymer carrier, which primarily provides good differentiation and adherence of osteoblasts under in vivo conditions, /7/.

The combination of HAP and bioresorbable polymers represents a new concept in the design of composite materials. This biomaterial can be considered as 'alive' because of its characteristic that allows it to gradually change its morphological properties in a previously programmed time period. The basic idea is that the bioresorption component (PLLA) gradually decomposes and makes places for the newly formed bone tissue. Studies have shown that the combination of HAP and PLLA results in the creation of a greater number of osteoblasts than the case with only HAP material (Fig. 2). As the number of osteoblasts increases, the reparation process is accelerated, /7/.

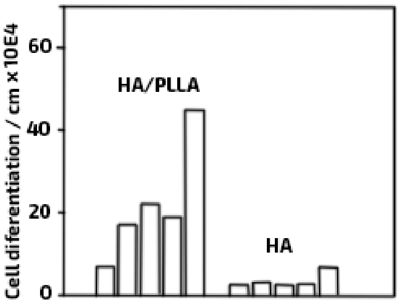

Figure 2. Increase in the number of cells on contact surfaces using PLLA polymers, /7/.

BONE TISSUE

Bones are the basic element of the skeletal system which together with the muscular system enables locomotor actions and provides mechanical protection to the internal organs. Bone tissue consists of organic and inorganic material. Organic material is collagen (90-95%). Collagen in the bones makes up about 40% of the total protein in the body and appears in the form of fibers. The remaining 5-10% is the inorganic component which is made of mineral salts. These salts have a crystalline form - they are hydroxyapatite structures. The crystals are predominantly made of calcium and phosphate, and significantly less calcium carbonate (10%). The percentage of mineral matter in bones depends on age, the functional stability of the kidneys, hormones, and enzymes, /8/.

It is expected that the hydroxyapatite composite will provide good biocompatibility in the bone-implant system because the hydroxyapatite is the basic inorganic component of the bone and has similar mechanical properties. Materials of the plates with high modulus of elasticity (titanium alloys) can lead to degradation of bone tissue beneath the plate.

NUMERICAL ANALYSIS

Different geometries of mandible fixation plates are created (Fig. 3). Numerical simulations and analysis of fixation plate shape influence on the distribution of stress at defined loads are performed. Some assumptions are introduced, both on the model and characteristics of the material.

During the development of the model it was necessary to introduce several assumptions. The first assumption is that the material is homogeneous and linearly elastic, whereas the friction between the contact surfaces is ignored. Contact between the upper jaw (maxilla) and lower jaw (mandible) is approximated by a flat surface at the front end of the jaw to achieve maximum stress at fracture. All tests are static, which is an approximation in comparison to real dynamic (fatigue) loads under *in vivo* conditions.

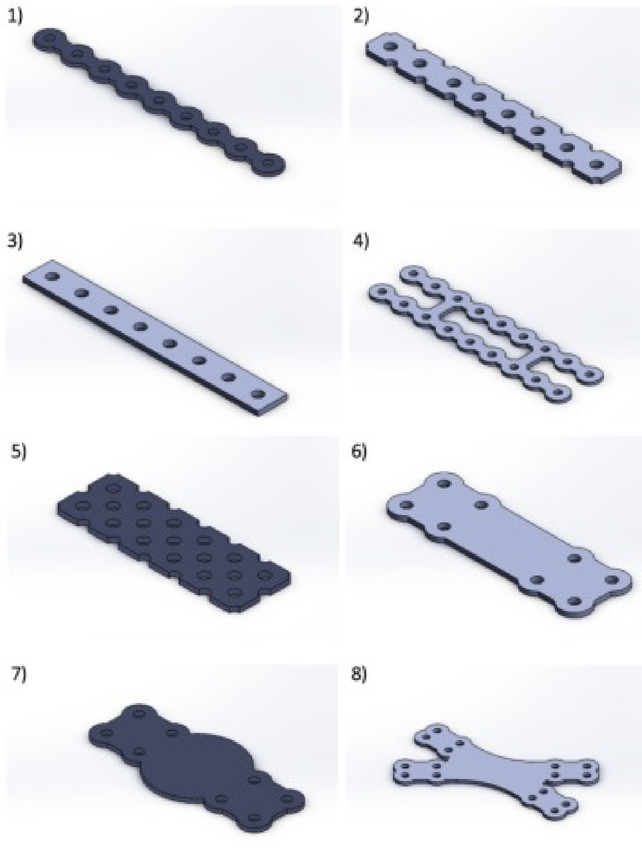

Figure 3. Examples of fixation plate models for testing.

The thickness of the mandible is constant and the fracture is flat as opposed to real fracture. The Poisson's coefficient for bone is 0.3 and the same value will be assigned to the tested HA structure - mechanical characteristics of HA are similar to bones, /9/. The modulus of elasticity is 20 GPa /2/, and yield stress is 184 MPa, /2/. The loading on mandible muscles is approximated by a force of 150 N /4/, whose direction is parallel to the surface of the plate.

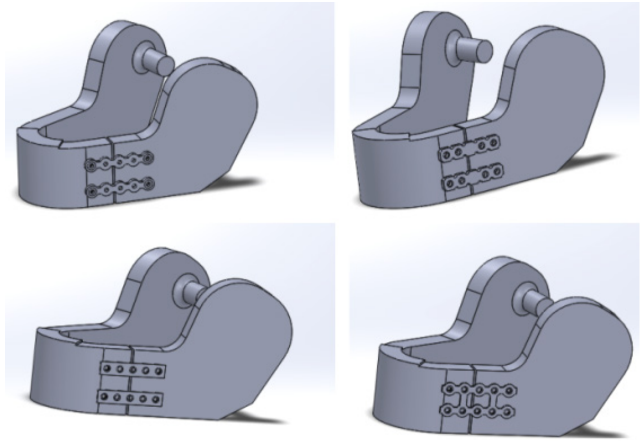

Figure 4. Model assemblies.

A total of eight models have been generated and the model was selected from references /10-13/. Some are in clinical use, while there are a few new models.

Figure 4 shows sets of the approximated mandible model and fixation plates. In cases of the plate model number 1, 2 and 3, another identical plate is placed to allow more accurate comparison with other plate design.

A large number of assumptions in the design of models will certainly lead to some inaccurate results. The contact between maxilla and mandible is approximated by a flat surface at the front end of the jaw to achieve maximum stress at fracture. The contact between cranium and mandible is approximated by a shaft with radius to avoid stress concentration. The wall thickness of the model is constant. The CATIA software package is used for the simulation.

The simulation needs to start from the basic assumptions of fixation. The contact area between the mandible and the maxilla is immovable when the occlusal forces reach their peaks (Fig. 5a). The shaft, which represents the contact of the cranium and the mandible, is fixed in two directions while rotation is possible (Fig. 5b). The value of the force is 150 N (total of 300 N) and represents the approximation of mandible muscles (Fig. 5c). The next step is to define the contact between mandible and fixation plates.

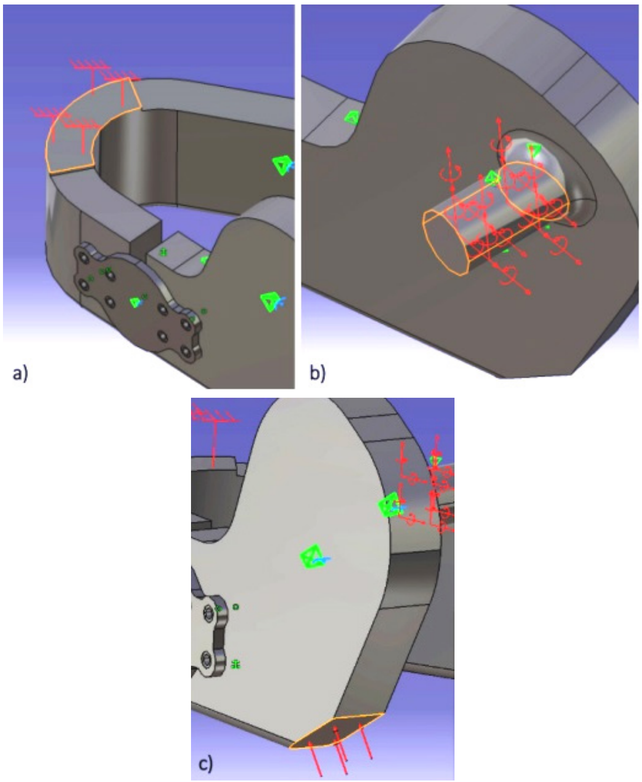

Figure 5. Boundary conditions and forces.

In Figure 6a, we can notice a direct contact of the plate and both parts of the mandible. The mandible is divided into two parts in order to perform the simulation. It is necessary to limit the movement of the openings both axially - screw head, and longitudinally - screw thread (Fig. 6b).

Numerical simulations are performed on eight fixation plate models.

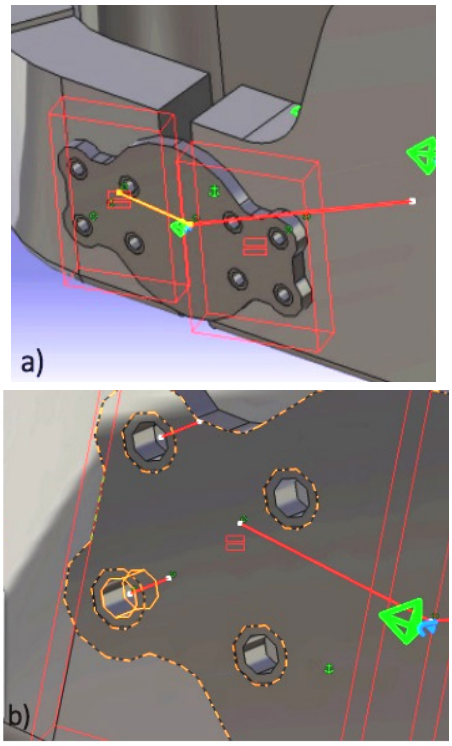

Figure 6. Fixation without penetration.

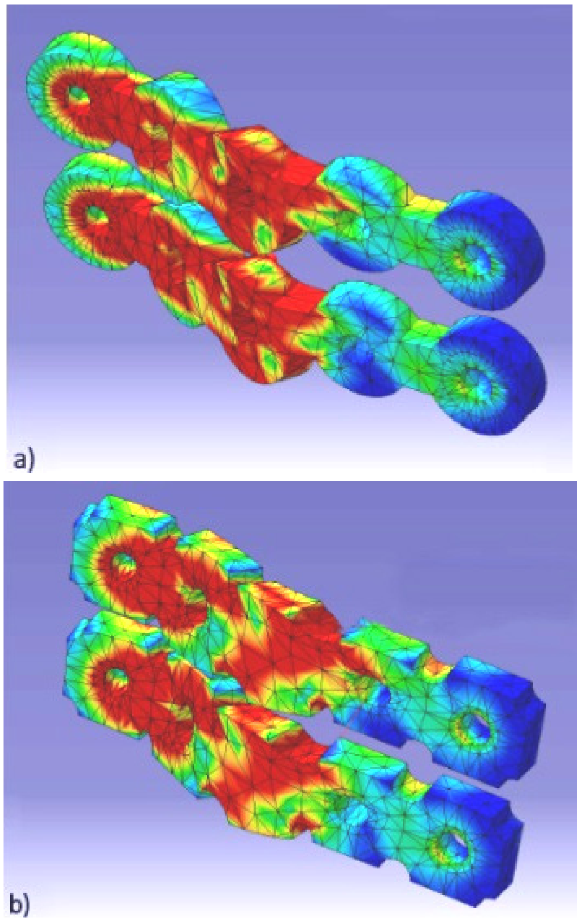

Figure 7. Fixation plates, models 1 and 2.

Figure 7a shows the maximum stress in model 1, which is about 35 MPa. When displaying results, a deformation scale is increased in order to make the changes visible. The maximum value is shown as red and is located mainly on the part of the plate near the fracture point. The maximum stress in model 2 is about 33 MPa, Fig. 7b. The maximum value is located on the part of the plate at the fracture point of the mandible.

Figure 8a shows the maximum stress in model 3, which is about 30 MPa. A noticeable stress drop is observed - up to 15% compared to model 1. The maximum stress of model 4 is presented in Fig. 8b. Unlike in previous tests, this model suffers significantly higher maximal stresses that reach about 50 MPa. The answer lies in the fact that the distance between the plates significantly affects the value of the maximum stress - the connection of the two plates does not lead to significant unloading. In this case the distance between the plates is shorter than in the case of model 1.

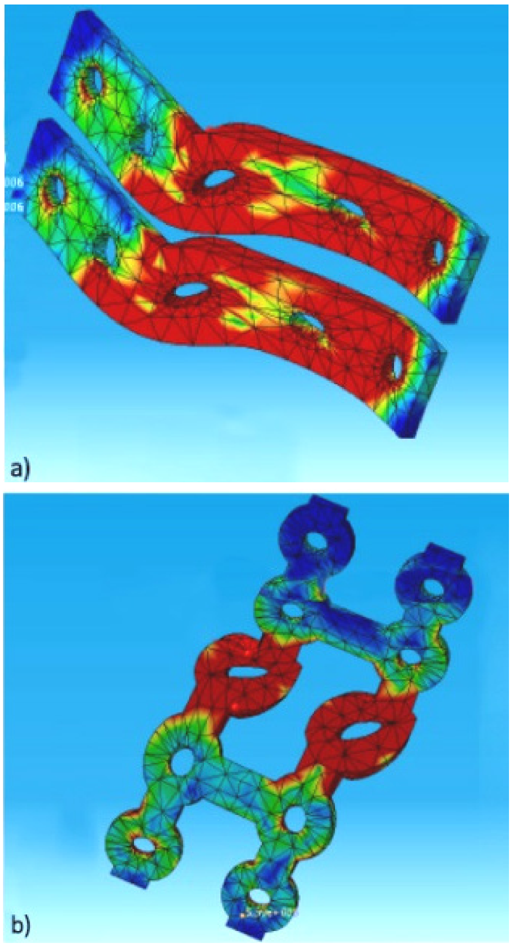

Figure 8. Fixation plates, models 3 and 4.

Figure 9a shows the maximum stress, which is about 34 MPa for model 5. This design of the plate had the idea of achieving a larger contact surface between the plate and the mandible - in previous cases we had spacing between the plates. Also, a number of openings have been drilled to enable the surgeon to fix the screws at his own will. The maximum stress in model 6 is about 20 MPa. This design of the plate had the idea of achieving the maximum contact surface without openings (Fig. 9b). The absence of openings has led to a better distribution of maximum stress by as much as 41%, compared to the separated plates (model 1).

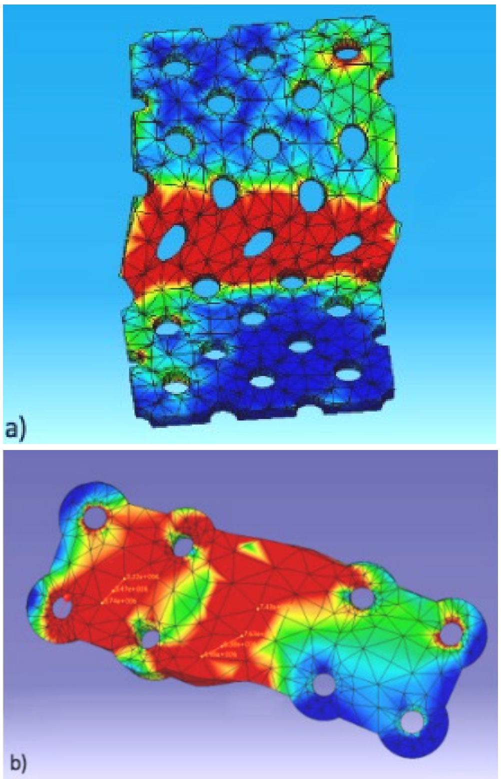

Figure 9. Fixation plates, models 5 and 6.

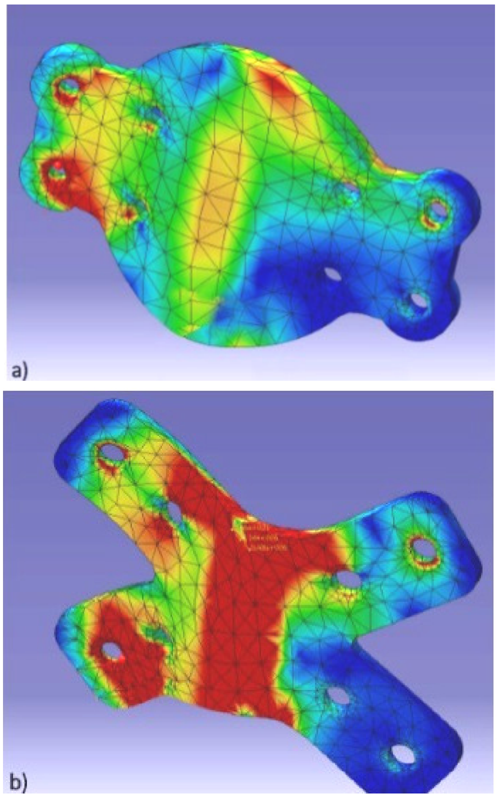

Figure 10. Fixation plates, models 7 and 8

In Figure 10a, the maximum stress in model 7 is estimated at about 9 MPa. This design of the fixation plate had the idea that the curvature from the side of the plate equally distributes the central load around the fracture. As we can see this is exactly what happened. The maximum stress zone is now concentrated predominantly around the opening closest to the effect of the force.

The maximum stress in model 8 is about 19 MPa, Fig. 10b. Unlike in the previous case, where the radius on the side has evenly distributed the stress, the stress in this case is predominantly centrally distributed.

ANALYSIS OF RESULTS AND DISCUSSIONS

From all of the above, it can be concluded that the fixation plate shown in numerical model 7 has a more even distribution of stress compared to the other plates. Its disadvantage is actually what gives it a good distribution of stress - a large surface. Naturally, the mandible is not an ideal flat surface as in this approximated model. This can lead to the loosening of screws, which later leads to failure of the implant.

Models that showed the best stress distribution characteristics are models 6 and 8. Results of maximum stress are almost identical, 19 and 20 MPa. The production of model 8 is slightly more complicated, and due to an identical stress distribution, we may concentrate our attention on model 6. Figure 7 shows a united graphic representation of maximum stress values for the tested fixation plate models.

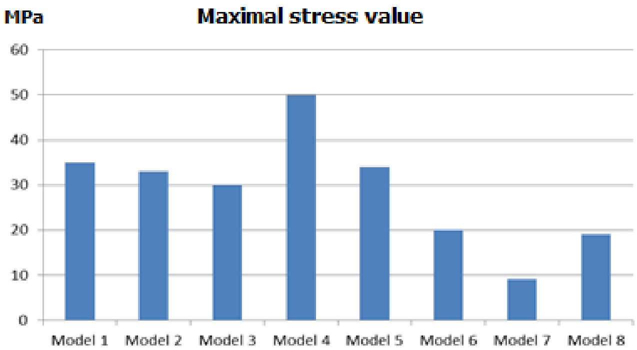

Figure 7. Maximum stress values for all tested models.

As previously said, this model is expected to show significantly better characteristics than any model of separated plates (models 1-3). The contact area of the mandible and fixation plate is relatively small compared to model 7, and because of this it can be expected that there will be no problems within the implant-bone system and there will be no loosening and potential failure of the implant. These types of plates would be relatively easy to fabricate, and due to the improved properties in relation to the separated plates, it should definitely be a design that needs to be tended.

CONCLUSION

Using the finite element method, the analysis of different forms of fixing plates is successfully presented. The obtained values of maximum tension on the plates should be taken carefully, because of making a few assumptions in the design of the model.

However, the accuracy in contrast between models is relevant, and so we can conclude that separate plates carry smaller loads but have a better fit to the bone. Plates with a larger surface achieve better stress distribution but there is a problem with contact area (naturally uneven) which can lead to loosening and potential failure of the implant. The proposed solution is presented as a fused plate with the smallest overall surface (model 6).

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