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# Stress–strain analysis of porous scaffolds made from titanium alloys synthesized via SLS method

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ABSTRACT

A layer-by-layer selective laser sintering (SLS) technology seems to be greatly promising for solving the plastic surgery problems, particularly those pertaining to the facial reconstruction. Made from titanium-based alloys (titanium or nitinol, i.e. NiTi-intermetallic phase), the porous scaffolds for cranioplasty are an efficient tool for rectifying the face defects and for the dental orthopedic surgery. The progress in the oral surgery and teeth implantation is caused by the problem of an osteointegration on the one hand, and by achievements of the implant synthesis techniques, on the other hand. An important problem thereby is a profound study of the stress–strain behavior of porous implants under the masticatory load or pressure. In the present study the ways for the optimization of the porous implant structural and strength properties as the function of the laser synthesis parameters are described. The finite element approach (ANSYS) was used here for a complex dowel description and numerical simulations.

In order to evaluate the processes in the porous implant under the external loading, a CAD 3D model was built for different internal and external configurations of the implant and/or initial shape of powdered particles. The stress–strain dependences were calculated that displayed the irregularity of the stress distribution by the implant volume in the bone tissue. Most of the values are concentrated in places of object contact.



## 1. Introduction

A perspective way of the implant synthesis is the process of a selective laser sintering (SLS) allowing the manufacturing of dental implants with open porosity and anatomically shaped scaffolds with a varying internal architecture. The porosity (permeability) and strength control can enhance a cell infiltration, stability of the nutrients mass transport and metabolic waste throughout the synthesized scaffold. Earlier we showed the possibility of the layer-by-layer laser synthesis of titanium and nitinol porous structures in order to confirm their suitability for the tissue engineering with a stable fixation and minimally invasive surgical procedures [1,2]. Nitinol is a NiTi-intermetallic phase and is of great interest due to its inherent shape-memory effect. The important problem here is to scrutinize the study of the stress–strain condition of the porous scaffolds under the influence of chewing and muscular loads on them. Also of interest is the optimization of the implant strength features and porosity in the CAD stage before the SLS process.

The present research is aimed at the verification of the synthesized implant structure on the basis of the powder particles shape, properties of the material used during the SLS and particles orientation in the spatial implant structure. The numerical modeling by the finite element (FE) method was carried out here, since the FE approach has been well developed and actively applied recently [3]. Achievement of the before-mentioned purposes will facilitate the manufacturing of the implants with a sufficient porosity for tissue ingrowth on the one hand, and satisfy chewing loads on the other hand. A numerical modeling of the porosity degree is expected to ensure the estimation of the redistribution of stresses within the contact area with the living bone tissue. One more object of interest is the influence of the model internal structure (shape and size of powder particles) and implant external shape on the resulting stresses, strains and displacements.

## 2. Model design and numerical approach

Firstly for the numerical simulations a simplified implant model of the cylindrical or complex shape was used. This construction was placed into the cavity of the presumably spongy bone tissue (Fig. 1). To shorten the simulation process, the model volume was quartered by symmetrically scaled meshing. Nevertheless, the boundary conditions were entered on the symmetrical model faces.

The implant porosity was determined via the structure of ordering objects (particles of the sintered powder) with a regular

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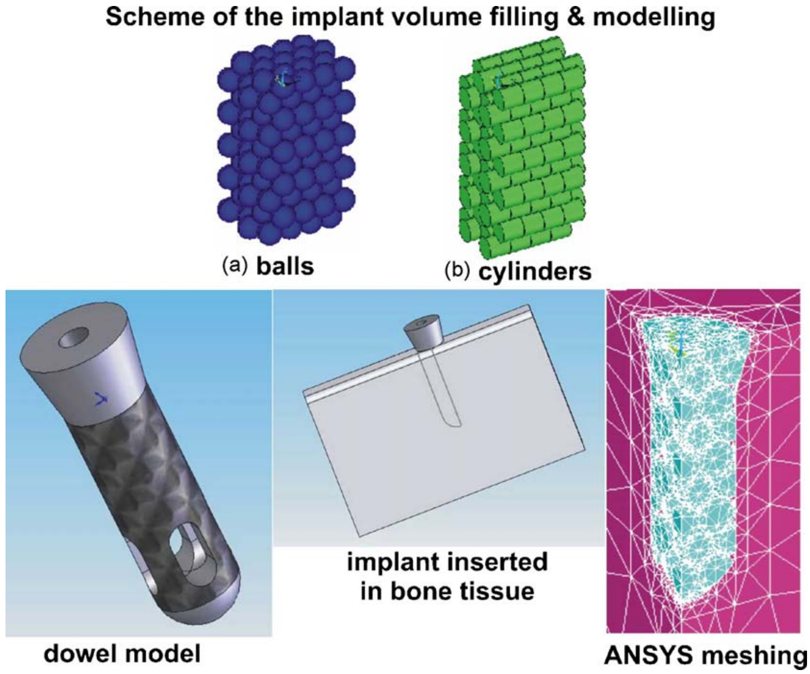


**Fig. 1.** Scheme of the implant volume filling and modeling. (a) balls; (b) cylinders.

shape (balls or cylinders were used, Fig. 1). The porosity value was adjusted for the ball diameter change. Also the porosity was described by changing a value of ball overlapping on its diameter, evaluated in percent. Practically, the degree of the ball overlapping is modeling of a laser sintering process. Material was designed as follows: one object (for example, ball) is created, then this object was copied several times along the $X$ axis on the distance of diameter overlapping. The duplicated process is repeated by $Y$ and $Z$ directions also. Herewith it was necessary to calculate that objects fell into the holes of close packed structure. Using the built-in ANSYS function (VINV) it was produced a cut-and-paste of model. The aggregate of balls (Fig. 1) was cut off by means of a three-dimensional cylinder (in the first case) or via a complex-shaped object (i.e. the dowel, in the second case), which were built in the Solid Works program. The size of these cylindrical or complex dowels was 1.5 cm $\times$ 2 cm. The spongy bone (Fig. 1) was presented as the surrounding volume, remotely attached at the distance of $\sim$6 cm. It was expected that this distance was enough to fix the bone.

We used 20-node SOLID95 element to describe the structure of titanium [4]. It allowed modeling of the transitional tasks and stress–strain behavior, as well as importing of the model from the CAD system. However, this element is not suitable for the material with a shape-memory effect. Thus, for nitinol we have chosen the SOLID187 3D-element with 10 nodes [4]. In the interaction (contact) zone between the implant and bone tissue, we concentrated the mesh of FE. The boundary of contact was described via TARGE170 and CONTA174 contact elements. The TARGE170 element was generated on the implant surface while the CONTA174 element was used on the internal surface of the bone cavity. It was an additional soft contact surface by our definition. We considered a yielding–yielding contact that implies a deformation of both materials. In our case, it was important to calculate the stresses both at the contact zone and in the bone tissue and porous metal part of the model.

A comparative analysis of the behaviors of the two types of titanium alloys was made. The Young modulus $E_X = 1.1025 \times 10^{11}$ Pa and Poisson ratio $\nu_{XY} = 0.32$ were selected for titanium and $E_X = 48 \times 10^9$ Pa, $\nu_{XY} = 0.33$ for nitinol (NiTi-phase). The spongy tissue parameters were $E_X = 6.89 \times 10^9$ Pa, $\nu_{XY} = 0.33$ [3,5].

It is known that the chewing load experienced by teeth is equal to 240 N [5]. So, the next boundary conditions were applied. Pressure was applied on the top external surface of implant, corresponding to this masticatory load. The bottom of bone surface was fastened, i.e. it was forbidden a displacement relative as $Z$-axis as the $X$–$Y$ axes, on the front part of it. The solution was produced in two steps of the ANSYS solver. At the first step of the solver, the implant "penetration" is achieved in the contact area of the spongy bone. This is caused by the displacement of the top surface layer along the $Z$-axis at the first step. The solution did not interrupt and the loading from the first step was kept. At the second step of the solver, the chewing load was effectuated. In the case of nitinol implant, the "Large Displacement Static"-solver was used [4] during the first step of loading; because of the "shape-memory effect" characteristic to this material.

During the first stage of our simulation we changed the ball diameter from 0.5 to 4.0 mm (Fig. 1a). Besides, we increased the value of the diameter overlapping from 2% to 9%. To make a comparison, we studied the two materials: titanium and nitinol. At the second stage of comparison, the model structure was represented in the cylindrical form of particles (Fig. 1b). Here we changed a structure of porosity as the percentage of the cylinder overlapping by their height. At the third stage of modeling, we studied the influence of the common implant shape introduced into the spongy bone construction. The shape of prosthesis (cylinder or dowel, Fig. 1) and its porous internal structure were designed. At last, the dependency of the simulation results on the loading direction was shown, i.e. how the result of simulation was changed under not only normal but also the tangential deformation comparable by their absolute values, applied on the upper implant surface.

## 3. Results and discussion

In Fig. 2 the results of the stress and displacement simulation are shown. The analysis of a solid-state FE model (i.e. without