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# Numerical analysis of unit cell models for orthopedic applications

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**Abstract.** For orthopedic applications 3D metal printing delivers flexible and tailor-made solutions. The greatest advantage of the additive manufacturing is that patient-specific implants can be manufactured. Determination of compressive properties of implant structure by numerical way is a demanding engineering task and it is indispensable for design purposes. In design of load-bearing biomedical implants the elastic behavior under working circumstances has to be considered. The professional literature offers different unit cell models to represent trabecular structures. In this paper the cubic open cell structures are established using CAD application and linear-elastic finite element simulations are performed. The aim of the research is to compare the material response of the investigated geometrical models.

## 1. Introduction

Foam structures produced by different technologies are widely spread in the engineering life. These types of cellular structures made new possibilities in different applications. It is well known that metal foam structures have high stiffness to weight ratio and excellent energy absorption capability. Several structures have been developed and investigated by researchers, e.g. open and closed cell metal foams [1-9] or syntactic metal foams [10-16]. Although metal foams are popular, they are still not sufficiently characterized thanks to its extremely complex structure which is usually stochastic in nature. Instead of modeling the complex internal structure directly researchers are using idealized structural approaches, e.g. unit cell model, statistical models, etc.

Additive manufacturing has also decades history. 30 years ago this technology using photopolymers let inventors to test their products as a prototype without investing high amount of energy in manufacturing. Nowadays metal printing is also available which was even a greater breakthrough to produce products without any major barriers.

The experience in the usage of unit cell models to represent metal foams have turned to design unit cell model based structures produced with additive manufacturing. There are number of suggestions for unit cell models worked out by researchers and engineers to represent cellular structures [17-19]. Numerous applications rely on the compressive property of lightweight cellular structures, which directly depend on its structure. As a load-bearing structural element (e.g. vehicle part, biomedical implant) lightweight cellular structure is expected to behave elastically under working circumstances, so the material response must be predicted precisely in the elastic region. Numerical determination of compressive properties of lightweight cellular structure is a forward-looking engineering task, and it is indispensable for design purposes.



The design process of biomedical implants is regulated to ensure the safety of patients. The directives and regulations connected to biomedical products are summarized and the design process is introduced in [20]. Although biocompatible materials have been used for decades as solid implants, they are limited by lack of fusion and bone resorption due to stress shielding. This results from the greater metal stiffness relative to trabecular bone [21], thereby the importance and necessity of different cellular structures to be developed as implants are increased.

In this paper the so called cubic unit cell model is analyzed numerically. According to the medical recommendations the cubic unit cell model is applied and established in CAD environment resulting different volume fraction values. Considering the design regulations the material response is predicted in the elastic region with the usage of finite element analysis to determine the maximum applicable load and the related compression value. The different geometrical variations are compared. The results of the numerical calculations can be applied in the design and stressing process of structures produced with additive technology.

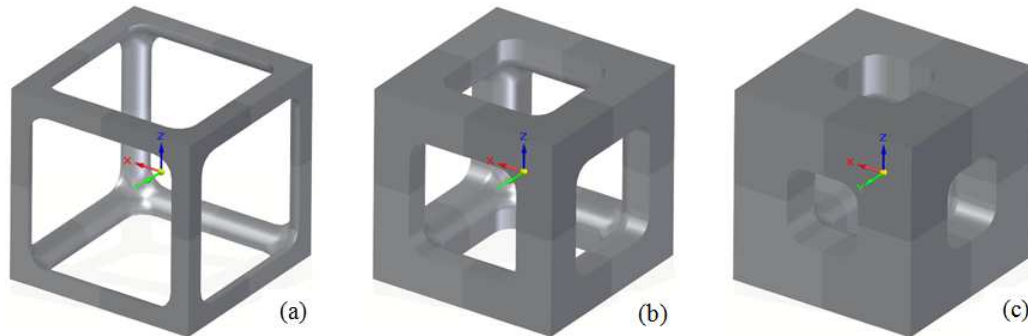
## 2. Materials and methods

For biomedical purposes special and biocompatible alloys are used, e.g. titanium alloys, tantalum alloys. The material and geometry selection is adjusted to Direct Metal Laser Sintering (DMLS) systems. With the usage of DMLS technology complex geometries can be formed accurately. The structures that are cannot be produced by other metal manufacturing process allows greater freedom for designers during the product development. Considering the finite element investigations, the relevant physical and mechanical properties and the chemical composition of the alloys and pure titanium are collected in Table 1.

**Table 1.** The properties of titanium alloys and pure titanium [22,23].

	Ti64 - Ti6Al4V	Ti64ELI		TiCP grade 2	
Material composition (wt %)					
Al	5.5 - 6.75	5.5 - 6.5		-	
V	3.5 - 4.5	3.5 - 4.5		-	
O	0.2	0.11		0.25	
N	0.05	0.04		0.03	
C	0.08	0.08		0.08	
H	0.015	0.012		0.015	
Fe	0.3	0.25		0.3	
Y	0.005	0.005		-	
Other, each	≤0.1	≤0.1		-	
Other, total	≤ 0.4	≤ 0.4		-	
Ti	balanced	balanced		99.325	
Physical property					
Density (g·cm <sup>-3</sup> )	4.41	4.41		4.59	
Mechanical properties					
	Heat treated	As built	Heat treated	As built	Heat treated
Ultimate strength (MPa)	1075	1290±80	1070±80	660	570
Yield strength (MPa)	965	1150±80	1010±80	560	445
Elastic modulus (GPa)	114.5		113.8		105
Poisson's ratio	0.34		0.342		0.37

The shape of the investigated unit cell models are cube with edge length 500  $\mu\text{m}$ . The edge length was determined from the medical recommendations that the size of the hole should be between 200 to 400  $\mu\text{m}$ . The geometrical model of the cubic unit cells are created in Solid Edge ST10, see in Figure 1.



**Figure 1.** Cubic models established in CAD environment: (a) 400R0-05, (b) 300R0-05, (c) 200R0-05.

The volume fraction of the unit cells is an important measure for describing cellular structures [24]:

$$\Phi(\%) = \frac{V_P}{V} \cdot 100 = \frac{V - V_S}{V} \cdot 100 \quad (1)$$

where  $\Phi$  is the volume fraction of pore phase,  $V$  is the total volume of the unit cell,  $V_S$  is the volume of the solid phase and  $V_P$  is the volume of the pore phase (the holes). The geometrical properties of the established cubic unit cell models are collected in Table 2.

**Table 2.** Geometrical properties of the unit cell models.

	400R0-05	300R0-05	200R0-05
Hole size ( $\mu\text{m} \cdot \mu\text{m}$ )	400·400	300·300	200·200
Round (mm)	0.05	0.05	0.05
$V_S$ ( $\text{mm}^3$ )	0.011	0.043	0.081
$\Phi$ (%)	91.2	65.6	35.2

The finite element calculations with Femap 9.3 was used to determine the load-displacement curves in the elastic region. The numerical investigations was based on the following equation:

$$\sigma_{eqmax} \leq \sigma_{all} = \frac{R_{p0.2}}{n} \quad (2)$$

where  $\sigma_{eqmax}$  is the maximum Von Mises equivalent stress,  $\sigma_{all}$  is the allowable stress for the material,  $R_{p0.2}$  is the yield strength and  $n$  is the safety factor.

The linear theory of the finite element method is used to determine the mechanical response in the elastic region of the investigated body:

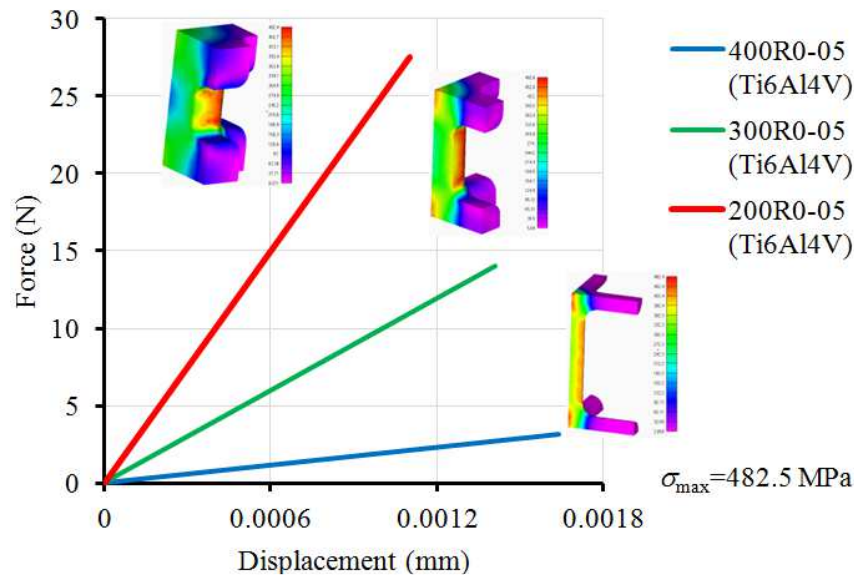
$$\mathbf{K}\mathbf{q} = \mathbf{f} \quad (3)$$

where  $\mathbf{K}$  is the global stiffness matrix, which refers to the property of the elements used,  $\mathbf{q}$  is the global nodal displacement vector and  $\mathbf{f}$  is the global load vector.

### 3. Results and discussion

The computational models were prepared to represent compression test and each consisted of a unit cell model and two rigid plates. Frictionless contact was assumed. The rigid top plate had a prescribed displacement and the bottom plate was used to measure the reaction force. The models were meshed

with tetrahedron elements and Ti6Al4V was used as material. The computational models were analysed using the commercial finite element software Femap 9.3 with NX Nastran solver for quasi-static loading. The applied safety factor is  $n = 2$  which is widely used in the orthopedic clinical practice. The force-displacement curves for the elastic sector are presented in Figure 2.



**Figure 2.** Force-displacement curves of the computational models.

The efficiency of each model compared to the bulk unit cell model is also determined where the efficiency is calculated in the following form:

$$E(\%) = \frac{F_{CMmax}}{F_{max}} \cdot 100 \quad (4)$$

where  $E$  is the efficiency,  $F_{CMmax}$  is the applied maximum force for the computational model and  $F_{max}$  is the applied maximum force for the bulk unit cell model. The results are collected in Table 3.

**Table 3.** The efficiency of the unit cell models compared to the bulk unit cell model.

	400R0-05	300R0-05	200R0-05
$F_{CMmax}$ (N)	3.14	14.02	27.55
$E$ (%)	2.60	11.63	22.84

The material response for 400R0-05 computational model was also determined using different materials with the same boundary conditions. For the calculations the applied safety factor was  $n = 2$ . The results of the numerical calculations are collected in Table 4.

**Table 4.** Material response for 400R0-05 computational model.

	Ti64 - Ti6Al4V		Ti64ELI		TiCP grade 2	
	Heat treated	As built	Heat treated	As built	Heat treated	
$F_{CMmax}$ (N)	3.14	3.75	3.29	1.83	1.45	
Displacement (mm)	0.00164	0.00196	0.00173	0.00104	0.00083	
$\sigma_{max}$ (MPa)	482.5	575	505	280	222.5	

#### 4. Conclusion

In the framework of the study numerical analyses were conducted to clarify the elastic behaviour of the so called cubic unit cell model with different geometrical properties. The established computational models were compressed by the usage of finite element software and force-displacement curves were analysed. A deviation occurred in the material response of different computational models, which originated in the actual geometrical properties. The development and characterization of cellular model variations allow stiffness selections to realize better match with modulus values of different bone types.

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