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DESIGN OPTIMIZATION OF DENTAL IMPLANT USING ADDITIVELY MANUFACTURED LATTICE MATERIALS.

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Abstract: *A Dental implant is a biocompatible surgical component placed into the jawbone to support dental prosthesis including bridges, crowns, or denture replacements. It might also be used in facial prosthesis operations or orthodontic anchoring. Currently, dental implants are constructed employing solid materials, coated with biocompatible layers. Since bone is a living tissue that is constantly modified in response to external loading, redistributed or reduced mechanical loading might cause bone resorption, implant loosening or interface failure, all of which have been notable problems for orthopedic implants. To address these issues, this paper presents a novel design for dental implant employing lattice materials. A lattice material is a class of open cell engineered cellular solid that is periodically structured and optimized for different applications. A multiscale and multi-objective design optimization framework based on Finite Element Method was developed to, primarily, minimize the bone/implant interface failure and bone loss and secondarily, to minimize the implant weight. Here, we assumed the implant as made of a lattice part (interface zone with the bone) and a solid part (implant core). The design variables included the microscopic parameters of the lattice unit cell as well as thickness of the interface zone. Simulation results show that the proposed design is capable of reducing the interface failure and the bone loss. Additively manufactured Titanium Ti-6Al-4V, a biocompatible material, is used for the new implant manufacturing, eliminating the need for a biocompatible coating.*

1 INTRODUCTION

Development of an ideal substitute for missing teeth has been one of the long-term aims of dentistry. A dental implant is a biocompatible screw-like ‘fixture’ that is surgically placed into the jawbone to replace the root of the natural tooth. The long-term benefits of dental implants include improved appearance, comfort, speech and self-esteem. With the dental implant, the patient can eat more conveniently and the inconvenience of embarrassment caused by removable partial and full dentures can be eliminated [1]. Major causes of implant failure are due to insufficient biomechanical bonding between the implant and the surrounding jawbone and also implant tooth fixtures or abutment failure[2]. Another key factor for the success or failure of a dental implant is the manner in which stresses are transferred to the surrounding bone. Load transfer from implants to surrounding bone depends on the type of loading, the bone-implant interface, the length and diameter of the implants, the shape and characteristics of the implant surface, the prosthesis type, and the quantity and quality of the surrounding bone [3].

Analyzing force transfer at the bone-implant interface is also an essential step in the overall analysis of loading, which determines the success or failure of an implant. Overload can cause bone resorption or fatigue failure of the implant, whereas under-loading of the bone may lead to disuse atrophy and subsequent bone loss [4].

In 1993 Clift, Fisher and Watson [5] studied the stress and strain distributions in the bone surrounding a new dental implant. Their design showed 50 percent reduction in stress concentration which would help to reduce fatigue failure and bone resorption in this area under lateral loading. Recently, an increasing amount of research has focused on the biological and mechanical behavior of highly porous structures of metallic biomaterials as implant materials for dental implants. Particularly, pure titanium and its alloys are typically used due to their outstanding mechanical and biological properties. However, these materials have high stiffness (Young's modulus) in comparison to that of the host bone, which necessitates careful implant design to ensure appropriate distribution of stresses to the adjoining bone, to avoid stress-shielding or overloading, both of which lead to bone resorption [6]. Several attempts have been undertaken to develop biomaterials with mechanical properties well suited to the bone tissue. Most of these studies have aimed at optimizing the important features of interactions between the implant surface and bone tissue. Modifying the implant surface can also improve the implant to bone interaction like Plasma spraying with different powder particles such as titanium oxide that has been used to coat dental implants [6-7]. Since bone is a live tissue which is continuously modified by the bone cells in response to external signals, reduced mechanical loading leads to resorption of bone, implant loosening and ultimately failure that has particularly been a problem for orthopedic implants in the past [7]. Numerous studies have been done to manipulate the mechanical and topographical properties of titanium implants. In many studies, micro and nano porous titanium has been proposed as a promising alternative to solid structures for biomedical and dental implant applications. Porous metals and metallic foams have combinations of properties that cannot be obtained with dense polymers, metals and ceramics or polymer and ceramic foams. For example, the mechanical strength, stiffness and energy absorption of metallic foams are much higher than those of polymer foams [8]. Many fabrication methods have been used to fabricate porous titanium for medical purposes. However, the size, shape, percentage and distribution of pores were variable and need further optimization [9]. One approach to overcome these drawbacks is using cellular structures or lattice materials which can provide a suitable biological environment for the host tissue to grow into the pores [10]. Lattice material is a microstructure made up of a regular repeating array of simple structural unit cell. Lattice materials have certain attractive features, such as the fact that the properties can in principle be predicted and characterised with a greater accuracy than when the pores are randomly located, though they have formerly been difficult to produce in anything other than elementary forms and small sizes, barring certain structures such as honeycombs [11].

Li and et al.[12] have studied the feasibility and evaluated the compressive properties of Ti6Al4V implants with controlled porosity via electron beam melting process. They found that the compressive yield strength of the Ti6Al4V implants with the porosity of around 51 percent is higher than that of human cortical bone while its Young's modulus is found to be similar to that of cortical bone. So, the porosities and mechanical properties of porous Ti6Al4V implants can be adjusted by changing porous structures, such as strut and pore sizes. Jamshidnia and et al. [13] also used electron beam melting method to produce bio-compatible dental implant designed by using non-stochastic porosity. They studied three different lattice structures including cross, honeycomb, and octahedral structures with different unit cell sizes to produce lattice abutment made of Ti6Al4V. Their investigations showed that the octahedral lattice structure with 2 mm unit cell size has the best mechanical behavior under 400 N normal biting force. Grunsven [14] used different strut thicknesses to produce a diamond lattice structure with graded porosity. In this study, it was found that the mechanical properties achieved could be relevant to orthopedic implants. Ahmadi et al. [15] presented new analytical solutions and closed-form relationships for predicting the elastic modulus, Poisson's ratio, critical buckling load, and yield stress of cellular structures made of the diamond lattice unit cell. They compared their results with experimental observations. According to their findings, there was a good agreement between the analytical predictions and experimental observations. Wang and McDowell [16,17] examined idealized individual cell wall behavior and determined the mechanical properties of the cell by solving deformation and equilibrium problems. Arabnejad and Pasini [16] studied asymptotic homogenization (AH) as a benchmark to test the accuracy of alternative schemes of homogenization applied to lattice materials. They applied AH to determine the effective elastic moduli and yield strength of six lattice topologies for a range of relative densities. They also introduced a methodology based on multiscale mechanics and design optimization to synthesize a graded cellular hip implant consisting of a lattice microstructure with nonhomogeneous distribution of material properties that can minimize concurrently bone resorption and implant interface failure [17].

In this study, a Finite Element Analysis (FEA) is presented to evaluate a new design of dental implant. It is assumed that dental implant is made out of a solid core and a lattice part in order to minimize the interface failure and bone loss of surrounding bone.

The effective mechanical properties of the lattice are predicted for octet-truss lattice and then used as the input for FEA model. The design variables of this study are the lattice zone thickness and the density of the lattice structure. The aim of this study is to determine the optimum interface thickness and lattice property in order to minimize the objective functions which, are the interface failure and bone loss.

2 COMPUTATIONAL MODEL DEVELOPMENT

In this section a 3-D CAD model of a segment of jaw bone and implant was developed, then the model imported into Altair Optistruct Hyperworks finite element software[18] to perform a nonlinear static stress analysis. The boundary conditions were also applied and then the properties of the lattice zone were found and used as the material properties in FEA.

2.1 GEOMETRY AND FINITE ELEMENTS MODELING

In this study, a 3-D model of a mandibular section of bone with 21 mm height and 20 mm width, representing the section of the mandible in the second premolar region, was modeled. As shown in Fig. 1, the model consisted of a spongy core, called cancellous bone, surrounded by 2.5 mm thickness of cortical bone envelop. On the other hand, the implant is considered as a single-piece part with 15 mm length and diameter of 4 mm in the cylindrical part, as shown in Fig. 2. The cylindrical section of the implant is assumed to be the part that forms the interface zone with the bone. This is the part of the implant that is considered for the lattice structure design. Mechanical properties and densities of the materials modelled in this study are tabulated in Table 1.

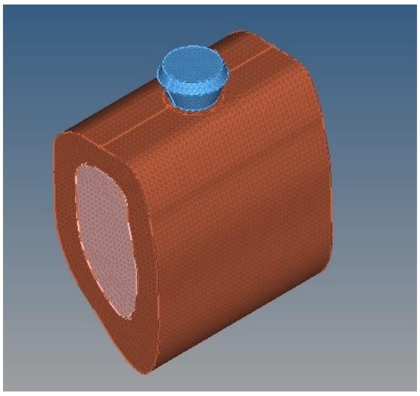

Figure 1: FEA model of bone and implant.

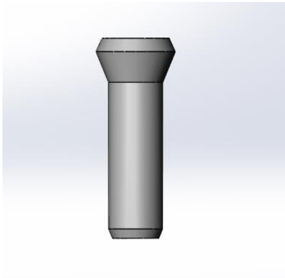

Figure 2: CAD model of implant.

Table 1. Mechanical properties of materials used in this study.

Material	Elastic Modulus (GPa)	Poisson ratio, υ	Density (gr/cm^3)	Yield stress (MPa)
Solid Titanium	103.4	0.35	4.5	880
Cortical Bone	13.7	0.3	2.17	_
Cancellous Bone	1.37	0.3	1.0	_

The 3D CAD model, then imported into Altair Optistruct Hyperworks finite element software in order to perform the nonlinear contact stress analysis. In Altair Optistruct the cylindrical part of the implant is divided into two zones. The solid core that is surrounded by the zone that interfaces with the bone and is made of lattice structure. The thickness of the interface zone as well as the lattice properties are the design variable. The contact surfaces were defined between the bone and the implant in order to simulate the bonding between them and it is assumed that there is perfect bonding between the bone and the implant. The thickness of the interface zone changes from 0.2 mm to 2 mm and the relative density of the lattice part is allowed to vary from 0.12 to 0.5. The lowest value of the relative density of the lattice material is selected so that the Young's modulus of the lattice material would be at least equal to or higher than the lowest Young's modulus of the bone.

2.2 BOUNDARY CONDITIONS AND CRITICAL LOADING

The implant was loaded with forces of 17.1 N, 114.6 N and 23.4 N in the lingual, axial, and mesiodistal directions, respectively [19], as shown in Fig. 3. These loads were determined by the work of Mericske-Stern and et al. [20]. They measured the force transmission onto implants in vivo by means of piezoelectric transducers and then measured the maximum loads for different type of denture anchorage. The bone segment is modelled as fixed at both ends. So the displacements of nodes at both ends in all directions are equal to zero.

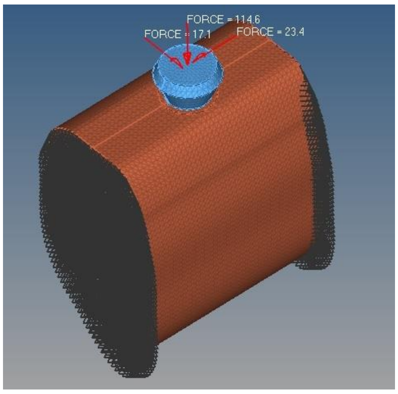

Figure 3: Applied loads and boundary conditions of FEM model.

2.3 EFFECTIVE PROPERTIES OF THE LATTICE MATERIAL

As mentioned, the new design is composed of a solid part which is the implant core and the lattice part which presents the interface zone between implant core and the surrounding bone. It is assumed that the interface zone is made out of lattice material. Relative density of the lattice material as well as the thickness of the interface zone are considered as the design variables in this study[21]. In this work, lattice material with regular octet-truss cell topology, as shown in Fig. 4, is selected to tessellate the interface zone of the implant.

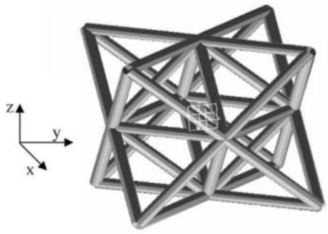

Figure 4: Structure of a unit cell of the regular octet-truss lattice material[22].

The relative density of the octet-truss lattice materials can be expressed as[22]:

$$\bar{\rho} = \frac{\rho_{La}}{\rho_0} = (6\pi\sqrt{2})(\frac{a_e}{l_e})^2 \qquad (1)$$

Where a_e is the radius of the cell element cross section, l_e is the element length, ρ_{La} is lattice material density, ρ_0 is the density of the solid material used to create the lattice material.

The relative Young's modulus of the octet-truss lattice material, $\bar{E}_{La}$, can be expressed as [22]:

$$\bar{E}_{La} = \frac{E_{La}}{E_0} = \frac{2\pi\sqrt{2}}{3}(\frac{a_e}{l_e})^2 \qquad (2)$$

where E_{La} and E_0 are the Young's moduli of the lattice and the solid materials, respectively [22].

By combining Equations (1) and (2) the relative Young's modulus of the lattice can be expressed as:

$$\bar{E}_{La} = \frac{E_L}{E_0} = \frac{1}{9}\bar{\rho}_{La} \tag{3}$$

According to equation (3) the relative Young's modulus of the lattice is directly proportional to its relative density. So the relative density can be selected as one of the design variables to control the interface stiffness.

3 IMPLANT FAILURE

Current dental implants are mainly made out of solid materials. Although significant improvements have been achieved, but two main drawbacks are still recognized for them: Interface failure and bone loss. Interface failure happens when the implant fails to integrate into the bone and bone loss happens because of the mechanical mismatch between the bone and implant. In this study, a new design of implant is presented to overcome these failures and reduce the implant weight.

4 DESIGN OPTIMIATION

The objective functions for optimization in this study are minimizing the bone loss, interface failure and the implant weight. Bone loss is the amount of bone that is going to be under loaded in the presence of the implant and interface failure is the probability of the mechanical failure at the bone implant interface.
To estimate the bone loss around the implant, the amount of bone that is under loaded can be assessed. Bone can be considered locally under loaded when its local strain energy (U_i) per unit of bone mass (ρ), averaged over n loading cases is beneath the local reference value S_{ref} , which is the value of S when there is no implant and is expressed as:

$$S = \frac{1}{n}\sum_{i=1}^{n}\frac{U_i}{\rho} \tag{4}$$

However not all under loading leads to bone loss [23]. In fact, bone resorption happens when the local value of S is beneath the value of $(1-s)S_{ref}$ [24]. The value of the dead zone, s, is assumed to be 0.5 in this expression. So the resorbed bone mass fraction can be expressed as:

$$m_r(b) = \frac{1}{M}\int_V g(S(d))\rho dV \tag{5}$$

where M and V are the original bone mass and volume in bone interface and $g(S(b))$ is the resorptive function which is equal to 1 when $S<(1-s)S_{ref}$ and equal to zero if $S>(1-s)S_{ref}$ [24].
To define the bone loss around the implant the number of elements that are unloaded can be counted and used to calculate the bone loss fraction.

The other objective function is the probability of local interface failure which can be expressed by the following function[24]:

$$F(b) = \int_\Omega f(\sigma)d\Omega \tag{6}$$

where $F(b)$ is the global interface failure function index, σ is the interface stress depending on the design variable, Ω is the interface area and can be defined by multi-axial Hoffman failure criterion [25]:

$$f(\sigma)=\frac{1}{S_t S_c}\sigma_n^2+(\frac{1}{S_t}-\frac{1}{S_c})\sigma_n+\frac{1}{S_s^2}\tau^2 \quad (7)$$

where S_t and S_c are the uniaxial tensile and compressive strengths, respectively, S_s is the shear strength, and σ_n and τ are normal and shear stresses at the bone-implant interface. If $f(\sigma)>>1$, a high probability of failure is expected, and if $f(\sigma)<<1$ the risk of interface failure is low. S_t, S_c and S_s can be expressed as a function of bone density as follows [26]:

$$S_t=14.5\rho^{1.71}, S_c=32.4\rho^{1.85}, S_s=21.6\rho^{1.65} \quad (8)$$

So the optimization problem can be formulated as:

$$Minimize: \begin{cases} m_r(b) & Bone\ Loss \\ F(b) & Interface\ Failure \\ m & implant\ mass \end{cases} \quad (9)$$

$$Subject\ to \begin{cases} \bar{\rho}\geq 0.12\ (Lattice\ relative\ density) \\ 0.2<t<2\ (Interface\ zone\ thickness) \\ \sigma_{VonMises(Lattice)}\leq\frac{1}{6}\bar{\rho}\sigma_{yield(Ti)}\ To\ avoid\ material\ failure \\ \bar{\rho}\geq\frac{6\sqrt{2}}{\pi}(\frac{\sigma_{yield(Ti)}}{E_{Ti}})\ To\ avoid\ buckling \end{cases} \quad (10)$$

Where *b* is the design variables vector which includes the interface zone thickness and the relative density of lattice. $\sigma_{yield(Ti)}$ and E_{Ti} are Titanium yield strength and young's modulus respectively and can be found in Table 1.

5 RESULTS AND DISCUSSION

Finite element models for different thicknesses and different relative densities were developed and the FEA results of normal stress, shear stress and local strain energy densities were used to calculate the interface failure and bone loss for each model.
The interface thicknesses between 0.2-2 mm were considered. The results showed that at thicknesses greater than 1.2 mm, the bone loss and interface failure increase significantly. Therefore, thicknesses between 0.2-1.2 were selected to study the bone loss and interface failure in order to determine the optimum thickness and relative density.

5.1 Interface failure

To evaluate the interface failure function for each model, normal and shear stresses at bone and implant interface were found using FEA. These stresses were used to calculate the interface failure probability in Equation (7) and then failure probability index in Equation (6) was calculated for each model. The results are shown in Table 2.

Table 2. Interface failure index results for different relative densities and different interface thicknesses.

Thickness of lattice	Relative density								
	0.12	0.15	0.20	0.25	0.30	0.35	0.40	0.45	0.50
t=0.2	28.50	28.82	29.25	29.59	29.87	30.09	30.27	30.37	30.55
t=0.4	28.11	28.38	28.76	29.08	29.36	29.60	30.06	30.00	30.16
t=0.6	28.11	28.33	28.67	28.97	29.24	29.49	29.71	29.91	30.09
t=0.8	28.44	28.59	28.86	29.13	29.38	29.62	29.84	30.05	30.24
t=1.0	28.71	28.83	29.06	29.31	29.55	29.77	29.99	30.19	30.38
t=1.2	28.92	29.04	29.27	29.52	29.77	30.00	30.22	30.43	30.61

5.2 Bone loss

As already discussed in Section 3, the bone loss can be determined by considering the number of under loaded bone elements around the implant. To investigate the bone loss, the local strain energy of the elements in the presence of the implant were compared to the corresponding local strain energy of the model when there is no implant. Then the bone loss was calculated according to Equation (5).These results can be found in Table 3.

Table 3. Bone loss percentage results for different relative densities and different interface thicknesses.

Thickness of lattice	Relative density								
	0.12	0.15	0.20	0.25	0.30	0.35	0.40	0.45	0.50
t=0.2	1.0	10.2	8.2	7.0	6.4	6.1	5.8	5.5	5.2
t=0.4	9.9	9.5	8.4	7.1	6.0	5.4	4.9	4.6	4.3
t=0.6	8.2	7.8	6.7	5.9	4.8	3.9	2.9	2.5	2.0
t=0.8	5.6	5.4	4.8	3.9	2.9	1.9	1.4	0.9	0.6
t=1.0	4.7	3.9	3.5	2.8	2.2	1.6	0.9	0.6	0.3
t=1.2	4.0	2.9	2.4	1.8	1.4	0.9	0.5	0.3	0.2

Figure 5 and Figure 6 depict the results for interface failure index and bone loss percentage as a function of relative density and thickness of the implant interface which is assumed to be made out of lattice. The results of interface failure index in Table 2 show by increasing the relative density, the interface failure index increases. By increasing the thickness from 0.2mm to 0.6mm interface failure index decreases for every relative density. But for thicknesses higher than 0.6mm the interface failure index increases. So the minimum value for this function happens at a thickness 0.6mm and relative density of 0.12. At this point the mass reduction would be 28.84% in comparison to a solid implant model. According to Table 3, the minimum bone loss happens when the interface zone thickness is 1.2 mm and relative density is 0.5. At this point bone loss is 0.2% and it can be selected as the optimum point when only bone loss is the objective. At this point, the mass reduction would be 27.24% in comparison to a solid implant. Figure 7 shows the implant structure with minimum bone loss expectation with interface zone thickness1.2 mm and relative lattice density of 0.5.

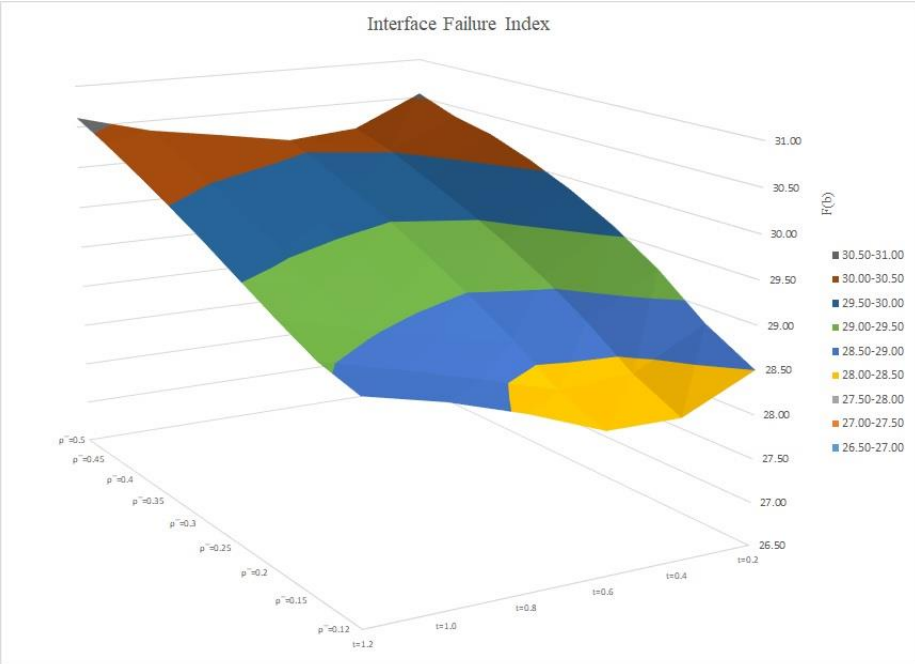

Figure 5: The effect of relative density and implant interface thickness at interface failure index.

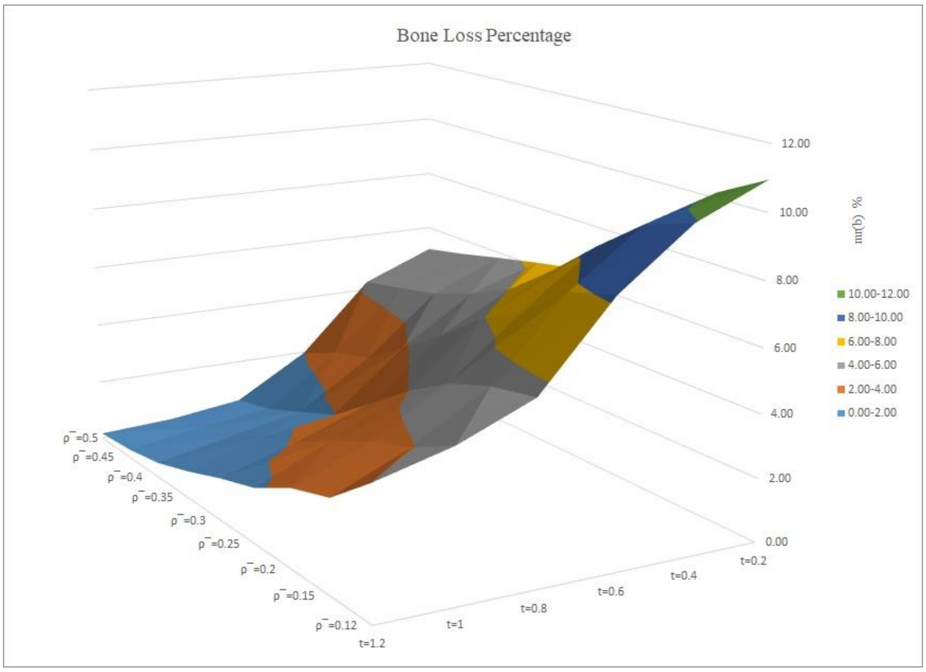

Figure 6: The effect of relative density and implant interface thickness at bone loss percentage.

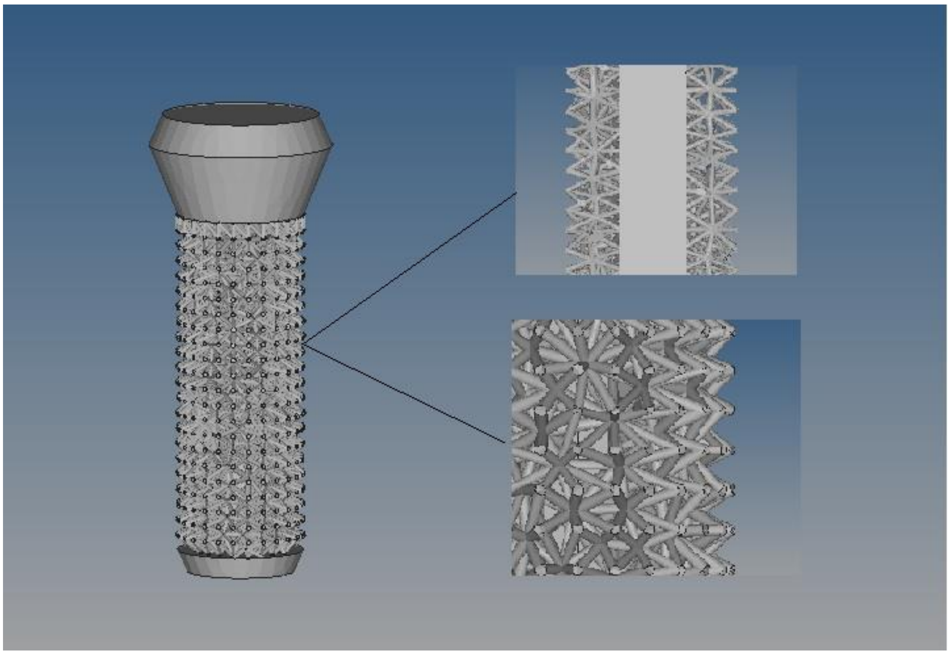

Figure 7: Implant structure after optimization

6 CONCLUSION

In this study, Finite Element Analysis was used to evaluate the effect of lattice structure in implant interface. The implant was considered as a cylinder-shaped model without the threads. The implant is made out of a solid core in cylindrical part and lattice part in interface zone. The octet-truss lattice was selected as the type of lattice unit cell and the interface thickness and lattice unit cell relative density were the design variables. Then the nonlinear static stress analysis was performed to determine the normal and shear stresses at bone implant interface. The results then used to find the local interface failure probability as well as the global interface failure functions. The results showed that by increasing the interface thickness from 0.2 mm to 0.6 mm the interface failure decreases. But by increasing the thickness from 0.6 mm to 2 mm, the interface failure index increases. So the optimum thickness in order to have the minimum interface failure index would be 0.6 mm for an implant with a diameter of 4 mm in the cylindrical part. To evaluate the bone loss, the strain energy densities of interface bone elements were studied and the results showed that by increasing the thickness and the relative density of the lattice part, the bone loss is decreasing and the minimum bone loss happens at thickness of 1.2 mm and % 50 of relative density. Using the lattice structure in implant design not only reduces the interface failure and the bone loss, it can also make the bone growth easier for the jaw bone. In comparison to solid implants, a lattice structure provides more space for bone to grow. So it can significantly decrease the bone loss around the implant. Also the final design will reduce the weight by 27-28%.

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