IOP Conf. Series: Materials Science and Engineering **324** (2018) 012060 doi:10.1088/1757-899X/324/1/012060

Structure Optimization of Porous Dental Implant Based on 3D Printing

Fangqiu Ji[1,2] ,Chunyu Zhang[2*] and Xianshuai Chen[2]

1 School of Mechanical Engineering, Shanghai University of Engineering Science, Shanghai, 201620, China
2 GuangZhou Janus Biotechnology Co., Itd. Floor 1, Buliding A, No.1121 Haibin Rd., Nansha, GuangZhou, 511458, China
E-mail: fq.ji@gzjanus.com, cy.zhang@gzjanus.com

Abstract. In this paper, selective laser melting (SLM) technology is used to process complex structures. In combination with the theory of biomedicine, a porous implant with a porous structure is designed to induce bone cell growth. The mechanical strength advantage of SLM was discussed by observing the metallographic structure of SLM specimen with mechanical microscope and mechanical tensile test. The osseointegration of porous implants was observed and analyzed by biological experiments. By establishing a mechanical model, the mechanical properties of the bone implant combined with the jaw bone were studied by the simple mechanical analysis under static multi loading and the finite element mechanical analysis. According to the experimental observation and mechanical research, the optimization suggestions for the structure design of the implant made by SLM technology were put forward.

1. Introduction

With the aging of the society, the number of people who need to repair dental teeth is becoming larger and larger. In addition to the traditional method of denture installation, more people have chosen a more stable and beautiful dental implant. At present, the production process of dental implant maturity is precision investment casting. There are many defects in the implants produced by this method, and the common problem is that the mechanical properties of the implants can not meet the standard because of the incomplete casting. Second, this method does not take into account individual differences. Human oral and maxillofacial bones are divided into four types of bone, each type of bone structure and density are different. The implant produced by this method can not accurately control the elastic modulus and other data. In the interaction with human oral and jaw, it will cause the problem of stress shielding, thus affecting the normal growth of bone [1].

2. Production principle and characteristics of SLM Technology

Selective laser melting technology has made up for the shortcomings of the original 3D printing technology, and made it develop rapidly in many industrial fields. The production principle of SLM technology is to draw 3D modeling of implants by computer aided design software such as CAD, and then slice the 3D graphics into slices by software. At the beginning of processing, the SLM equipment controls the powder laying system to lay a layer of metal powder in the molding cavity, and the scanning system controls the laser beam to be burned on the powder according to the image information of the first slice. After the first layer of powder is processed, the system controls the piston to move the worktable downward, which is equal to the distance of a layer of powder. Then a powder is laid on the first layer of the processing, and then the second layers are processed and formed. Repeat the above steps until the entire workpiece is formed.

SLM technology not only has fast processing speed, but also has high machining precision. SLM technology is a kind of material processing method, so it can save a lot of precious raw materials in the process of production, thereby reducing the cost greatly. The most outstanding feature of SLM is that it can make complex shape and internal structure according to the work needs, so as to achieve better performance requirements. Therefore, the porous structure of the implant can be printed by its 3D, thereby changing the elastic modulus of the implant, and ultimately avoiding the problem of stress shielding. The porous structure implant can further optimize the structure design according to the finite element analysis structure.

3. Performance analysis of SLM specimen

Using German equipment SLM125 YLR-100-W, the commercial two grade pure titanium (CP-Ti Grade 2) powder was used as the raw material to produce 3D test piece. Then, the metallographic structure is observed by metallographic microscope, as shown in Figure 1. The metallographic structure of the part is shown in Figure 2. The metallographic structure of the parts produced by the traditional casting process is shown. It is not difficult to find that the Figure 1 is very fine acicular martensite, indicating that it contains higher strength and hardness. Figure 2 shows that it contains more martensite, because the martensite microstructure is decomposed into fine acicular $\alpha+\beta$ phase when heated in the $\alpha+\beta$ region, and the phase is easy to fracture due to shear mechanism [2]. The mechanical properties of the parts produced by SLM technology are better than those produced by the traditional casting process. Through tensile test, the yield strength of SLM specimen is 654 MPa, while the yield strength of commercial pure titanium forgings is 590 MPa[3]. It is shown that the strength of SLM specimen is better than that of forging.

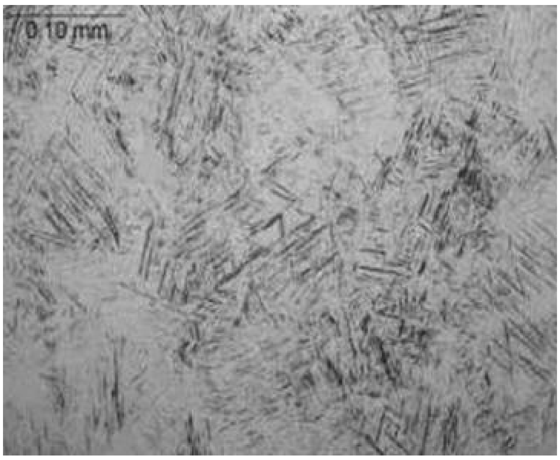

Figure1. metallographic structure of SLM workpiece Figure.

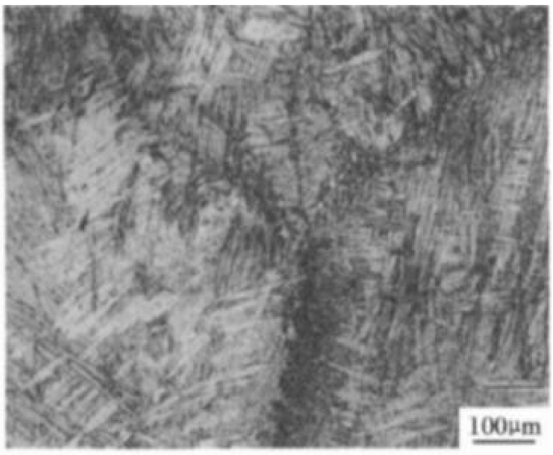

Figure2. metallographic structure of casting parts.

4. Design and experimental analysis of porous structure

In view of the biocompatibility of implant and human jaw, as well as to avoid stress shielding effect between the implant and the implant, and to improve the physical properties of implant. Based on the characteristics of SLM technology which can be used to manufacture complex structures, some porous structures are designed in this paper. Porous structure materials are widely used in biomedical materials because of their unique role in improving biocompatibility.

As shown in Figure 3, the implant is embedded in the outer part of the oral cavity and jaw using a porous structure design. Such a design is the use of porous structure can adhere to the growth of bone cells and promote bone cell ingrowth into the gap, so that the implant is more stable and jaw fixed. It

IOP Conf. Series: Materials Science and Engineering **324** (2018) 012060 doi:10.1088/1757-899X/324/1/012060

is beneficial for the stress of the implant to be dispersed into the jaw and reduce the stress of the implant and the denture on it. In addition, the porous structure can adjust the density, strength and elastic modulus of the material by adjusting the porosity, so that the mechanical properties of the implant and the jaw match, and finally achieve the purpose of weakening or eliminating the stress shielding effect. Because of the improper treatment of stress shielding effect, the necrosis of the bone near the implant and the severe consequence of reducing its bearing capacity [4].

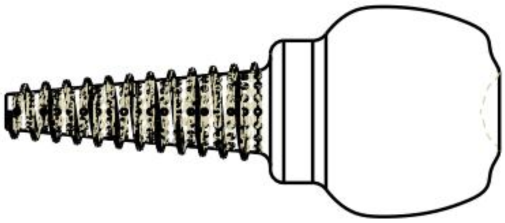

Figure 3. schematic diagram of implant structure.

At present, studies in the field of medicine show that the pore size of implants suitable for bone cell ingrowth is 100~600 μm, considering the self recovery of capillary and jaw tissue, so the pore size is larger than 300 μm[5]. Fukuda [6] found that the porous 500 μm implant had the best effect on the ingrowth of bone cells through the implantation of different pores on the animal skeleton. Therefore, the aperture design of 500 μm is reasonable, and the specific numerical value needs to be further determined. The laser spot of the equipment SLM125 is 87 μm. Theoretically, the manufacturing accuracy is enough to produce the implant according to the design requirements.
According to the existing medical research, the porosity of human cortical bone is 5.36% ~ 14.1%, the pore size is 5~150 μm, the compressive strength is 160 ~ 200MPa, and the elastic modulus is 3 ~ 30 GPa. The porosity of cancellous bone is 30% ~ 95%, the pore size is 20~1000 μm[7], the compressive strength is 4 ~ 7 MPa, and the elastic modulus is 3 ~ 30 GPa. According to the jaw structure of different individuals, the implant can increase its biocompatibility with the jaw by adjusting the porosity. Therefore, the optimal scheme is to design the porosity distribution of the implant bone segment into two segments according to the different porosity distribution of the implant cortical bone and cancellous bone. At the same time, the implant elastic modulus approximate to the human skeleton can also effectively avoid stress shielding [8]. The elastic modulus of porous structure can be obtained according to the [9] formula of Nielsen model:

$$E = E_0 \times \frac{(1-\varphi)^2}{1+\left(\frac{1}{\rho}-1\right)\times\varphi} \tag{1}$$

In the formula, E is the elastic modulus of porous structure, E_0 is the modulus of elasticity of compact structure, φ is porosity, and ρ is shape factor. Assuming that the designed implant is the standard optimal shape, that is, the shape does not affect the structural strength of the implant, then ρ =1 is simplified (1) into a finite element:

$$E = E_0 \times (1-\varphi)^2 \tag{2}$$

It is known that the elastic modulus of SLM compact specimen is 110GPa, and the elastic modulus of porous structure should be designed from 3 to 30GPa according to the principle of similar elastic modulus. Therefore, the design range of porosity is 48% - 84%, which is similar to the porosity of human cancellous bone. Although it can not meet the approximate requirement of human skin bone porosity, similar elastic modulus can still avoid stress shielding. Therefore, the porosity of the cortical bone segment in the implant was further optimized when the porosity and elastic modulus could not be optimized simultaneously.

According to the design range of the porosity, the exact porosity value was selected randomly, and the pore size parameters of 500 micron m were used, and the standard rotational structure of the implant was made by SLM. Biological experiments were carried out and planted in the thigh of New Zealand rabbits. After eight weeks, the implant was scanned by histological section and CT scan, and the 3D reconstruction of CT scan was performed by software. As shown in Figure 4, the microstructure of the implant was reconstructed by microscopic CT, and the yellow was the new bone tissue. It can be seen that the bone tissue has entered into the internal pores, and the osseointegration is good.

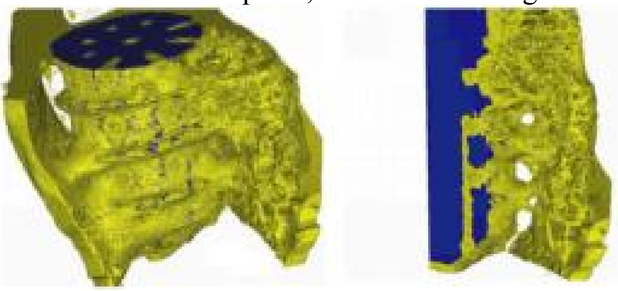

Figure 4. 3D CT reconstruction of implant tissue

5. Physical analysis of implant

5.1 establish the mechanical model

As shown in Figure 5, the anatomical structure of the dental implant after the implantation of the conventional dental implant is shown. Dental implant is composed of denture, implant foundation and abutment. Implant foundation is divided into cortical bone segment and spongy bone segment. Abutment is also called gingival segment. In the study, the design was used to simulate the force of the implant. Firstly, the mechanical analysis of the implant under the static multi loading condition was carried out. As shown in Figure 6, this is the schematic diagram of molar force simulation. In order to facilitate the calculation, it is simplified as shown in Figure 7. The length of the implant in the figure is a, and all the implants are implanted into the jaws. The length of the B part is the abutment of the implant, which connects the implant and the artificial denture. Kong and other [10] for long-term clinical trials, and from a biomechanical point of view, the diameter of the implant should be greater than 4.0 mm, length should be greater than 11.0 mm. According to the characteristics of Asian teeth, the preliminary design of abutment height B is equal to 3.0 mm, the artificial denture height h is 7.0 mm, and the pile foundation length a is 14 mm. The implant foundation is equivalent to a cylinder, the diameter of R_2 is 6.0 mm, and the implant abutment and the artificial denture are equivalent to a cylinder, and the diameter of the R_1 is 8.0 mm. The a segment is an implant foundation embedded entirely in the alveolar bone, which is considered to be surrounded by the homogeneous load of the alveolar bone. The B segment is embedded in the part of the periosteal flap, and its stress on the implant can be neglected due to its soft tissue characteristics.

When chewing the food, the molar plays a major role, which mainly bears the combined action of compressive stress, shear stress and torque. In the establishment of the mechanical model, it is assumed that the implant is in an ideal state, that is, the bone cell ingrowth into the implant and make it into an integral body with the jaw. Because the mechanical properties of the implant conform to the assumption of continuity, uniformity and isotropy, it is equivalent to the bar and then its mechanical analysis is carried out. By drawing the bending moment diagram of the implant, the dangerous section is determined as the surface L1 in figure 7. Studies have shown that normal people's molar will chew about 98 N to 225.4 N bite force [11], so F_1 take 225.4 N. Taking into account the masseter muscle plays a leading role in masseter muscle, and the masseter muscle is the control of the upper and lower jaw occlusion, so chewing teeth under the condition of the horizontal direction of the force F_2 is small, according to the oral physiology of Ohammed research, take F_2 120 N [12].

In this mechanical model, according to figure 7, the bending moment M1 is produced by the interaction of F_1 and F_2. According to the working characteristics of molars, the vertical load is regarded as uniform load, and its horizontal force is regarded as a point. Thus, the maximum torque and bending moment of the implant under the limit condition are obtained by using this model, and the formulas for calculating the torque M0 and the bending moment M1 are as follows:

$$M_0 = F_2 \times \frac{R_1}{2} \tag{3}$$

$$M_1 = \int_m^n \mathrm{qxdx} + F_2(b+h) \tag{4}$$

In the formula (m, n) for the F_1 interval, $q=F_1/R_1$, F_1 as a uniform load on the surface of the denture. By calculation, M_0 is equal to 0.48 N • m, and M_1 equals 1.43 N • m.

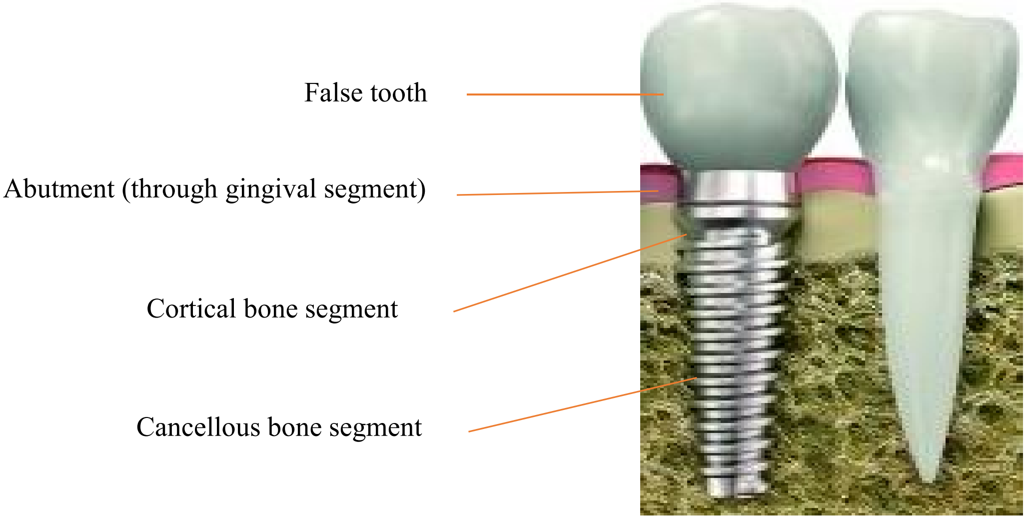

Figure 5. implant implant simulation

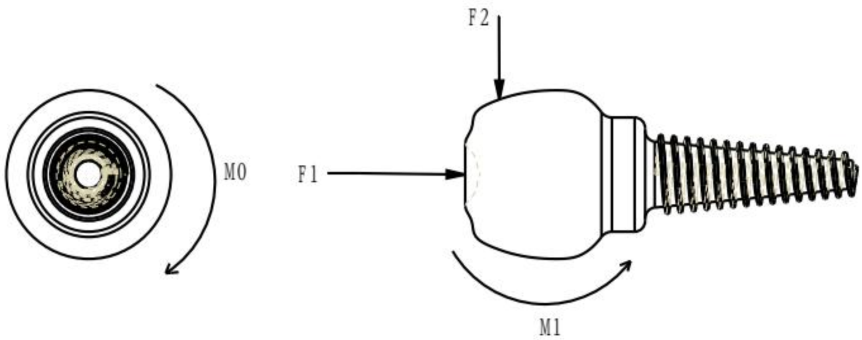

Figure 6. sketch of simulated force of molar

IOP Conf. Series: Materials Science and Engineering **324** (2018) 012060 doi:10.1088/1757-899X/324/1/012060

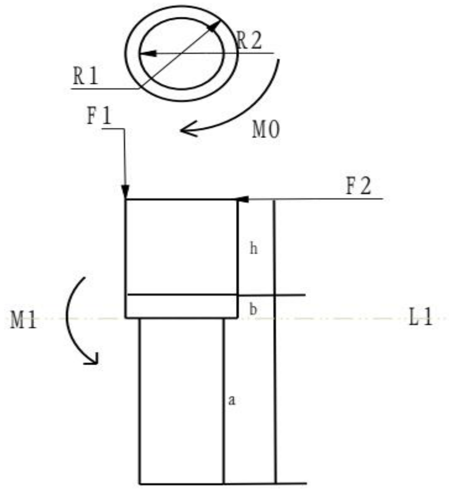

Figure 7. simplified model of molar force model

5.2 finite element analysis

The hypothesis of the mechanical model is that the implant and the jaw form a whole, ignoring the influence of the porous structure of the implant. However, the bending moment and torque are also present in the porous implant, and the data obtained from the model is similar to the true data in the porous structure when the implant and the jaw are in good osseointegration. After removing the simplified model of osseointegration, a finite element model for simulating the stress of implant with porous structure was established. The static multi loading mechanical model was calculated and the relevant data were imported into Ansys software to simulate the stress finite element analysis. Figure 8 is the deformation analysis diagram. According to the simulation results, the maximum deformation occurs at the abutment position of the implant. Figure 9 is the stress analysis diagram. According to the simulation results, the stress at the joint between the neck and the abutment of the pile is greatly influenced by the stress. The maximum stress of the implant at the dangerous interface is 35.1MPa, which is less than the yield strength of the compact structure SLM specimen. The relationship between the yield strength of porous structure and porosity and pore size remains to be experimentally investigated. The dangerous interface between the finite element and the mechanical model is similar to that of the dangerous interface, which indicates that the mechanical model is of higher accuracy. Through the stress distribution diagram, it is not difficult to find the bone graft part of the implant foundation, that is, the stress of the porous structure site is smaller, and less than 3.9 MPa. The experimental data will provide suggestions for further optimization of implant structure.

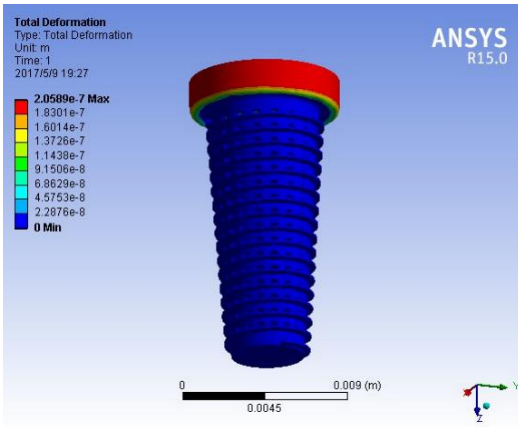

Figure 8. deformation analysis diagram

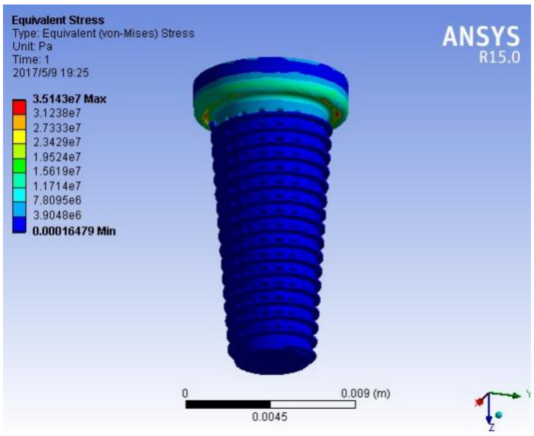

Figure 9. stress analysis diagram

6. Discussion

In this paper, the strength advantage of SLM technology for manufacturing implants was analyzed by observing the microstructure and tensile test of SLM pure titanium specimens. According to the theory of osseointegration, it is known that the porous structure contributes to the ingrowth of bone cells, and the porous structure can effectively avoid the problem of stress shielding. At the same time, controllable manufacturing of porous structure is another advantage of SLM technology compared with traditional manufacturing process.

According to the existing biomedical theories, it is demonstrated that the pores of 500 μm are the optimal size for inducing bone ingrowth. According to the machining accuracy of the existing German equipment SLM 125, the machining dimension of 500 μm porous structure pore is determined. In order to avoid stress shielding, it is necessary to adjust the elastic modulus of the implant close to the human bone tissue by adjusting the porosity of the implant. According to the Nielsen model and the elastic modulus distribution range of the human bone tissue, the design reference porosity of the implant was 48% - 84%. At the same time, the porosity of implant and jaw is beneficial to improve the biocompatibility of the implant, so the exact porosity value should be determined according to the specific porosity of the patient's jaw. The porosity distribution range of the implant is close to that of the cancellous bone, but the distribution of the porosity is different from that of the human skin. Therefore, the porosity meets the requirement of biocompatibility when the implant is implanted into the porous structure of cancellous bone segment. When designing the porous structure of cortical bone implants, only the design requirements of elastic modulus can be taken as the optimum design standard. Considering that the porosity of cortical bone is small, the elastic modulus of cortical bone is larger than that of Nielsen model. According to the reasons of stress shielding, it is known that the elastic modulus of implant is not greater than that of human bone tissue. Therefore, it is reasonable to design the porosity of implants larger than the porosity of cortical bone. Through biological experiments, the porous implant with pore size of 500 μm and porosity of 48% - 84% was verified to have good biocompatibility, and a better implant jaw bone bonding plane was formed.

The formation of a good bone bonding plane is the prerequisite to ensure the initial stability of the implant, and the implant itself has enough strength and stiffness is the key to ensure the service life of the implant. Through theoretical analysis, the size and length of dental implants were determined. According to the analysis of oral anatomy, the load of implant in the limit condition was determined. By establishing a static multi loading mechanical model after implant implantation, it is assumed that the implant and jaw form a whole, ignoring the influence of the porous structure of implant. The mechanical properties of the implant are consistent with the assumptions of continuity, uniformity, isotropy, and then equivalent to the bar. Then the complex three-dimensional force is converted to two dimensions, and the key data of the moment and torque are obtained. The location of the dangerous interface of dental implant under static multi loading was determined.

The static load moment and torque obtained by the mechanics model of the implant in the conventional finite element simulation will be ignored, but it is real, and had better osseointegration in implant and jaw, model data and real data obtained in the porous structure is near. At the same time, the bending moment and torque are helpful to simplify the bone bonding model and remove unnecessary complex parts such as denture. The osseointegration model of porous dental implant and alveolar bone was established, and then the finite element analysis was carried out. The results of finite element analysis is the design of the porous structure of the strength and stiffness of the implant with normal work safety requirements, and to determine the similar position and static risk interface multi load mechanics model of the position, further validation of the static load mechanical model of high precision. According to the finite element analysis, the deformation of the implant occurs at the neck and base of the abutment, which is due to the fact that the implant can not transmit the force directly to the jaw in the gingival segment, which is consistent with the hypothesis of the static multi loading mechanical model. Therefore, when designing the neck and abutment of the implant, the stiffness should be increased as much as possible in the design requirements, that is to reduce the porosity and improve the elastic modulus. At the same time, this design also meets the requirements mentioned above to eliminate stress shielding and improve biocompatibility to reduce the porosity and increase the elastic modulus of implant cortical bone segment. According to the stress analysis diagram, most of the stress is concentrated in the connecting section between the implant neck and abutment, and the dangerous interface is between the bone graft section and the gingival segment. Therefore, in the design of porous implant, the stress concentration should be avoided and the density should be increased at the connection between the implant neck and abutment. It also shows that the two - segment implant has better structural strength than the one - stage implant. The stress of the porous structure of the whole cancellous bone segment is less than that of the human cancellous bone. Therefore, according to the different bone conditions, the intensity of individual design of porous structure is in line with the design requirements.

7. Summary

Through the existing biomedical theory, it is proved that porous structure with pore size of 500 μ m and porosity of 48% ~ 84% is proved to be beneficial to induce bone regeneration, and the rationality of the derivation is proved by biological experiments. The design of porous structure for porous implant according to different bone conditions is in line with the design requirements, but the connection between the implant neck and abutment should be designed with high density to eliminate stress shielding and improve the strength and stiffness of implant.

Reference

[1] Traini T, Mangano C, Sammons RL, et al. Direct laser metal sintering as a new approach to fabrication of an isoelastic functionally graded material for manufacture of porous titaniumdental implants [J]. Dent Mater,2008, 24(11):1525-1533.

[2] Xin S W,Zhao Y Q． Discussion on heat treatment and precipitation phase of titanium alloy[J].Metal heat treatment,2006,31(6):39-42.

[3] X.P. Li,J. Van Humbeeck,J.P. Kruth. Selective laser melting of weak-textured commercially pure titanium with high strength and ductility: A study from laser power perspective[J]. Materials & Design,2017,116:.

[4] Van der Stok J, Van der Jagt OP, Amin Yavari S, et al. Selective laser melting-produced porous titanium scaffolds regenerate bone in critical size cortical bone defects [J]. J Orthop Res, 2013,31(5):792-799.

[5] Polo-Corrales L, Latorre-Eateves M, Ramires-Vick JE. Scaffold design for bone regeneration [J]. J Nanosci Nanotechnal,2014,8(1):15-56.

[6] Fukuda A, Takenoto M, et al. Osteoinducation of porous Ti implants with a channel structure fabricated by selective laser melting [J]. Acta Biomaterialia,2011,7(5):2327-2336.

[7] Dabrowski B, Swieszkowski W, Godlinski D, et al. Highly porous titanium scaffolds for orthopedic applications [J]. Biomed Mater Res-applbiomater,2010,95B(1):53-61.

[8] Tan B S. Effect of Biomechanics on long-term retention of implants[J]. Chinese Journal of Prosthodontics,2014,15(02):122.

[9] Kang J F,Wang L. Microstructure design of 3D printing variable modulus metal prosthesis[J]. Journal of Mechanical Engineering,2017,53(05):175-180.

[10] Kong L, Li T, Wang Y,et al. Selection of optimal dental implant diameter and length in type IV bone: a three-dimensional finite element analysis[J]. Oral Maxillofac Surg,2009,38(10):1077-1083.

[11] Guo N S,Liu J P. The stress analysis of periodontal membrane of fixed abutment tooth with different thickness［J］.Chinese Clinical rehabilitation 2004,29(8):6352-6353.

[12] Ohammed Ibrahim M, Thulasingam C, Nasser KS, et al. Evaluation of design parameters of dental implant shape , diameter and length on stress distribution:a finite element analysis[J]. Indian Prosthodont Soc,2011,11(3):165-171.

Acknowledgments

This work is supported by the Science and Technology Program of Nansha, China (No. 2016GJ009).