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Numerical investigation of the effect of porous titanium femoral prosthesis on bone remodeling

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ABSTRACT

Porous titanium is a promising orthopedic implant material. As a potential use in total hip replacement, the effect of a porous titanium femoral prosthesis on bone remodeling is investigated in this paper. The stress and strain fields of a post-operative femur with a hip replacement are calculated by applying the three-dimensional finite element method. The effect of the implant material on the bone remodeling is evaluated by analyzing the loss of bone density following a strain magnitude based bone remodeling theory. Different implant materials, including currently used solid cobalt–chrome and solid titanium, potential porous titanium with different porosities, are considered in this study. This investigation confirms that bone loss around the implant strongly depends on the value of the elastic modulus of the prosthesis. There will be a sharp drop of the volume of the bone with density loss if a cobalt–chrome implant is replaced by a porous titanium implant. The numerical results show that both of the bone volume with density loss and the bone density loss rate decrease linearly with the increase of the porosity. However, increasing porosity will reduce the strength of porous titanium. With regard to material design for porous titanium-based femoral prosthesis, stress analysis is required to meet the strength requirement.

1. Introduction

Hip replacement surgery was one of the most surgical advances in the last century. Since the first surgery in 1960, improvements in joint replacement techniques have greatly increased the effectiveness of total hip replacement. Nowadays, it was reported that about 600,000 hip replacements are performed worldwide every year and 90% of recipients are over 65 years old [1]. The femoral prostheses used for hip replacement are made from solid alloys, such as cobalt–chrome and titanium alloys. Following hip replacement, bone loss around femoral prostheses occurs due to the altered post-operative loading environment, particularly due to stress shielding. In some serious cases, bone loss can lead to the loosening of the prosthesis and repeat surgery may be needed [2].

Stress shielding is mainly caused due to the mismatch of the elastic modulus between the implant material and the bone. For example, the elastic modulus of cobalt–chrome alloy is 210 GPa while the elastic modulus of cortical bone is only about 17 GPa [3,4]. Since the implant is much stiffer than the bone, some of the forces applied on the bone have been shielded by the prosthesis, which results in bone density loss. A solution to this problem is to develop new implant materials with the stiffness closing to that of bone, such as composite materials [5] and porous titanium [6,7]. Besides the property of low elastic modulus, open-celled Ti alloy foams allow bone tissue growing in the implant materials [8,9].

To assist the development of porous titanium as implant materials for hip replacement, the effect of a femoral prosthesis made from porous titanium on bone remodeling was theoretically investigated in this study. The bone remodeling, i.e., here, bone density loss, was quantified by applying a strain magnitude based theory. The strain magnitude field was calculated from the three-dimensional finite element modeling of a post-operative femur under typical loading conditions. The three-dimensional finite element simulation plays an increasing role in the development and design of femoral prostheses [5,10,11].

2. Finite element model

2.1. Geometry of the femur

The three-dimensional geometry of a femur can be established from computed tomography (CT) images. The femur geometry varies from person to person and it changes with age even for the same person. From the research standpoint, to simplify the experimental cross-validation of numerical studies, researchers from Rizzoli Orthopaedic Institute, Bologna, Italy, established a database called International Society of Biomechanics Finite Element Mesh

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Repository on the internet. The so called "muscle standardized femur" [12] from the database was downloaded and utilized in this study.

2.2. Geometry of the implant

In the practice of total hip replacement, every implant has a specific geometry which is adapted for the fixation. In the current study, the geometry of the three-dimensional implant was developed based on the study of femur anthropometry [13] and the documentations of femoral prostheses [14,15].

2.3. Finite element meshes

The geometrical models of the femur and the implant were created by using the software UGS NX5. After that, the finite element meshes were generated by using the software MD R2 Patran. Two finite element meshes, one for an intact femur and one for a post-operative femur, were developed, which are shown in Fig. 1. For the study, it is important to have the same mesh for the part of the bone which is present before and after surgery. That will allow a comparison of stress and strain in the same element in both cases.

The finite element calculations were carried out by applying the software Abaqus. Both second order and first order tetrahedral elements were used. Convergence was checked before finalizing the meshes. As shown in Fig. 1, both of the intact and post-operative femur consist of a large number of elements. The total number of finite element nodes for the post-operative model is 918,473 and it took about 8.5 h to finish a calculation in a supercomputer SGI Altix 2700 Bx2 cluster.

2.4. Material data

Bone occurs in two forms: as a dense solid, cortical bone, and as a porous network of connecting rods or plates, cancellous bone. In the literature of numerical simulations, both cortical bone and cancellous bone were usually considered as a homogeneous, isotropic and linearly elastic material, e.g., Turner et al. [16]. Rohlman et al. [17] have shown that the stress distribution in the bone is influenced only slightly when anisotropic material is assumed. Moreover, the exact material behavior of the bone varies individually

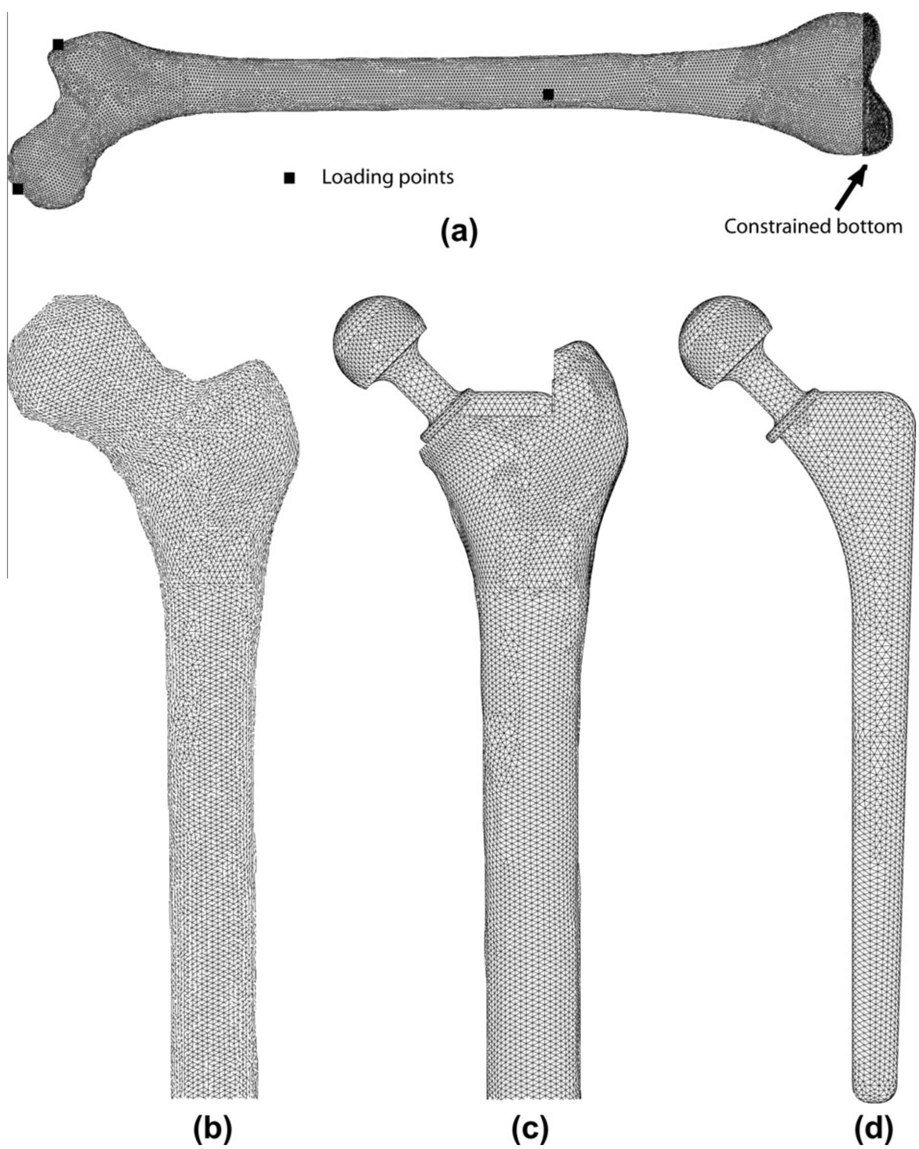

Fig. 1. Finite element meshes: (a) the mesh for the entire intact femur; (b) the partial mesh of the intact femur with femoral head; (c) the partial mesh of the post-operative femur with femoral head; (d) the mesh of the femoral prosthesis.

Table 1
Material data used in the FE calculations.

Material	Young's modulus (GPa)	Poisson's ratio
Cortical bone	17	0.33
Cancellous bone	0.4	0.33
Cobalt–chrome	210	0.3
Titanium	110	0.3
Titanium foam with 20% porosity	70.4	0.3
Titanium foam with 30% porosity	53.9	0.3
Titanium foam with 40% porosity	39.6	0.3
Titanium foam with 50% porosity	27.5	0.3
Titanium foam with 60% porosity	17.6	0.3

and locally, so it would be very difficult to have a perfect description of the anisotropy of the bone. In the current study, cortical bone and cancellous bone were assumed as homogenous, isotropic and linearly elastic. Their elastic constants were taken from Taylor et al. [4] and shown in Table 1.

Cobalt–chrome alloy and titanium alloy are currently applied to produce femoral prosthesis. Cobalt–chrome alloy is much stiffer than titanium alloy. Their elastic moduli and Poisson's ratios are listed in Table 1. With regard to the new material of porous titanium, its properties depend on the relative density of the foam ρ^*, which is related to its porosity p_t by

$$\rho^* = \frac{\rho_f}{\rho_a} = 1 - p_t \tag{1}$$

where ρ_f is the density of the foam and ρ_a is the density of the titanium solid.

According to the study of Gibson and Ashby [18], the elastic modulus of an alloy foam, E_f, and its the yield strength σ_{Yf}, can be predicted by

$$E_f = C_1 E_a (\rho^*)^2 \tag{2}$$

$$\sigma_{Yf} = C_2 \sigma_{Ya} (\rho^*)^{3/2} \tag{3}$$

where E_a and σ_{Ya} are respectively the elastic modulus and yield strength of the solid alloy. C_1 and C_2 are constants, and a value of 1.0 for C_1 and 0.3 for C_2 were suggested. Some researches indicate that predicted results from Eqs. (2) and (3) agree well with experimental data from porous titanium [9,19]. Therefore, these formulas were applied in the study to predict the elastic modulus and yield strength for titanium foams with different porosities. The porosity does not influence significantly the Poisson's ratio and a value of 0.3 is frequently used.

2.5. Loading conditions

Loads on a femur are applied by muscles and joint contact forces. There are 31 muscles connected to a femur [12]. The load magnitude and direction from each muscle depend on the activity. It is very difficult to know all the values for each activity phase from each muscle. Simplifications are usually made to consider major muscle forces under critical situations in the numerical study of the mechanics of femur [3]. In this study, the loading model from Ramos et al. [20] was adopted. This model was defined for the most strenuous phase of a typical walking cycle. In this model, there are three concentrated forces applied on the center of the muscle insertion area. The same forces were applied to all the intact and post-operative femurs. The locations of the three forces are indicated in Fig. 1a. The bottom of the femur, which connects to the knee, is constrained in all the simulations, as shown in Fig. 1a.

3. Bone remodeling theory

Change in the physiological loading will lead to the change of bone density, i.e., bone remodeling. After total hip replacement, the loss of bone density around the prosthesis is expected. To evaluate the effect of implant material on the loss of bone density, a strain-adaptive remodeling theory is applied. This theory was originally proposed by Huiskes et al. [21] and then modified by Turner et al. [16]. According to this theory, the bone remodeling rate is defined as the bone density change rate $d\rho/dt$, which is determined by [16]:

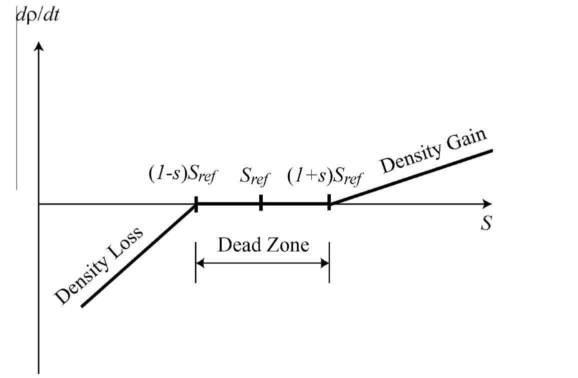

Fig. 2. Illustration of bone density change rate as a function of the magnitude of the strain tensor.

$$\frac{d\rho}{dt} = \begin{cases} aC_{Inc}[S - (1+s)S_{ref}] & \text{when } S > (1+s)S_{ref} \\ 0 & \text{when } (1-s)S_{ref} \leqslant S \leqslant (1+s)S_{ref} \\ aC_{dec}[S - (1-s)S_{ref}] & \text{when } S < (1-s)S_{ref} \end{cases} \tag{4}$$

within the limits of $0.1 < \rho < 2.0$ g/cm^3, where $S = \sqrt{\varepsilon_{ij}\varepsilon_{ij}}$ is the magnitude of the strain tensor and a is the surface area density. C_{inc} and C_{dec} are the coefficients of the density increase and decrease rate, respectively. Based on clinical remodeling rates, Turner et al. [16] specified the density decrease coefficient rate C_{dec} to be 3.5 times greater than the density increase coefficient C_{inc}, which is applied in the current study. The surface area density a depends on the porosity of the bone p_b and it can be estimated by [22]:

$$a(p_b) = 32.3p_b - 93.9p_b^2 + 134p_b^3 - 101p_b^4 + 28.8p_b^5 \tag{5}$$

This bone remodeling theory can be illustrated in Fig. 2. There are three zones in the $d\rho/dt$ versus S diagram, a density gain zone corresponding to positive $d\rho/dt$, a density loss zone corresponding to negative $d\rho/dt$ and a dead zone with zero $d\rho/dt$. The width of the dead zone depends on the parameter s, which is taken as 0.60 [16]. As confirmed in our numerical study, density gain is not expected after total hip replacement.

4. Results and discussion

4.1. Bone volume with density loss

A set of finite element calculations were carried out for an intact femur and several post-operative femurs with different implant materials. The purpose of simulating an intact femur is to obtain the strain magnitude tensor at each element, which is then used as the local value of S_{ref} in the bone remodeling theory, Eq. (4), to evaluate bone density loss in a post-operative femur. Applying the bone remodeling theory, as shown in Fig. 2, we can find the total volume of the bone with density loss in a post-operative femur by using the following criterion in each element:

$$\text{If } \frac{S}{S_{ref}} < (1-s), \text{then the bone in the element will lose density} \tag{6}$$

Clearly, the calculated volume of bone with density loss depends on the loading condition, which affects the local reference

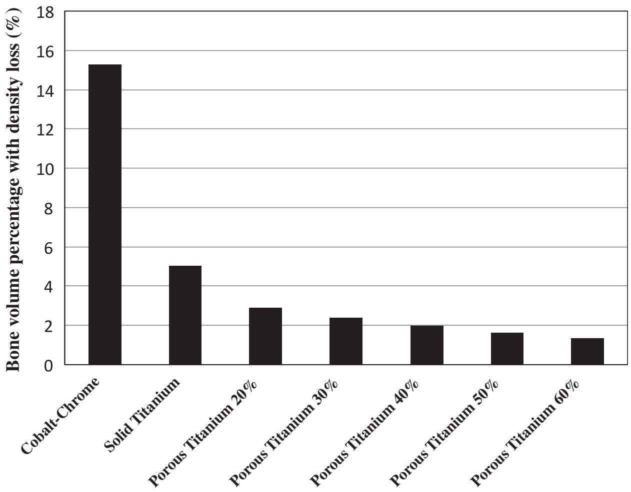

Fig. 3. The calculated volume of cortical bone with density loss for different implant materials.

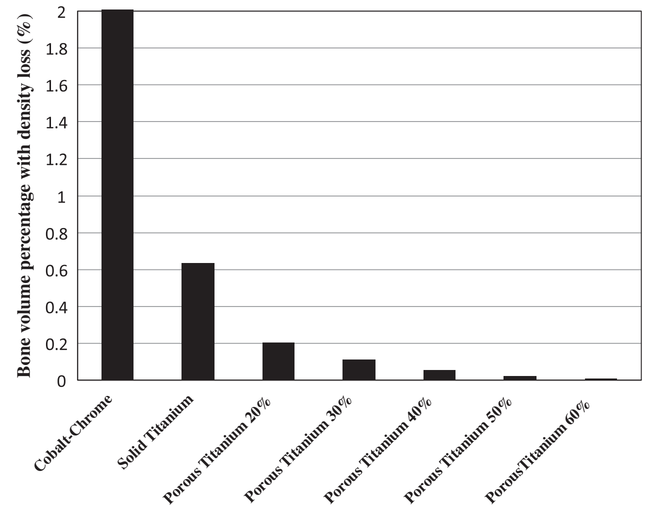

Fig. 4. The calculated volume percentage of the cancellous bone with density loss for different implant materials.

value of S_{ref}. Under the loading condition in the most strenuous phase of a typical walking cycle specified in previous section and $s = 0.6$, Fig. 3 shows the bone volume percentage with density loss in the cortical bone for different implant materials. Loss of bone density after total hip replacement was reported in clinic research. For example, loss of bone density ranging from minimal, mild to severe was observed in a set of 57 total hip replacements over 6 years of study [23]. It is worth mentioning that titanium alloy was the implant material in the published study [23]. Fig. 3 further indicates that if the implant is made of cobalt–chrome, the calculated volume percentage with density loss in cortical bone is about 15% (the total volume of the cortical bone is 205,309 mm^3). Because the predicted value depends on the loading condition and choice of s value, the comparison of the performances of different implant materials would be more reliable and relevant. As we can see from Fig. 3, there will be a significant drop of the volume percentage with density loss in the cortical bone if the cobalt–chrome implant is replaced by the solid titanium implant. The volume of the cortical bone with density loss is three times lower in the case of solid titanium implant. This large difference in the volume should be due to the large difference in elastic modulus ($E_{cobalt-chrome} - E_{titanium} = 100$ GPa). As we know, bone density loss is mainly caused by stress shielding. Reducing the mismatch of the elastic modulus between the implant and the bone should mitigate the problem of stress shielding, therefore, reducing bone density loss. This finding is consistent with previous numerical studies, e.g., [24,25]. Our finite element mesh is much finer than those presented in [24,25]. Additionally, the current study focuses on the potential use of porous titanium as implant materials.

Fig. 3 also shows that the cortical bone volume with density loss can be reduced further if porous titanium materials are applied as the implant material and the volume decreases gradually with the increase of the porosity of the titanium foam. From the solid titanium to the titanium foam with 60% porosity, the volume of the cortical bone with density loss is expected to drop about 73%. This volume decrease is also believed due to the decrease in the elastic modulus from solid titanium to porous titanium, $E_{solid\ titanium} - E_{titanium\ foam\ 60\%} = 92.4$ GPa. According to Fig. 3, less than 3% of the total cortical bone in all the porous titanium cases will lose density. For solid titanium, it is close to 5%.

Fig. 4 shows the volume percentage of the cancellous bone with density loss for different implant materials. The results are similar as those for the cortical bone. A sharp drop between the cobalt–chrome and the solid titanium implants and a gradual reduction of the volume with the increase of the titanium foam porosity can be observed. But it appears that there is a very small volume of the cancellous bone which is in loss of density (less than 0.6%) with solid titanium or porous titanium. Moreover, the fixation is principally on the cortical bone. Therefore, we will only focus on the density loss in the cortical bone in the following discussions.

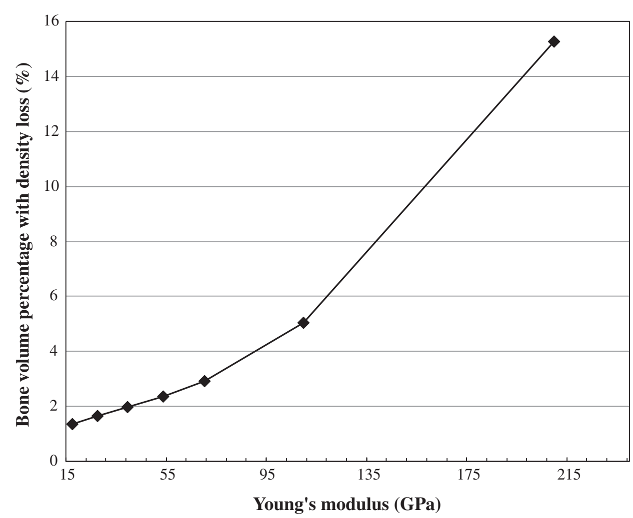

Fig. 5. The calculated volume percentage of cortical bone with density loss as a function of the implant's elastic modulus.

As bone density loss is related to the elastic modulus of the implant material, a quantitative relationship from the calculated results is plotted in Fig. 5. This graph confirms that the volume of bone with density loss depends on the elastic modulus value of the implant. When the elastic modulus of the implant and of the bone are close ($E_{cortical} = 17$ GPa), the volume is low. These results explain why an implant made of porous titanium can reduce bone loss.

From the finite element simulations, we cannot only calculate the total volume of the bone with density loss but also predict the locations where bone loses density. Fig. 6 shows the predicted results in cortical bone for different implant materials. Bone density loss occurs around the fixation area near the head of the femur in all the cases, which agrees with the results by Turner et al. [16].

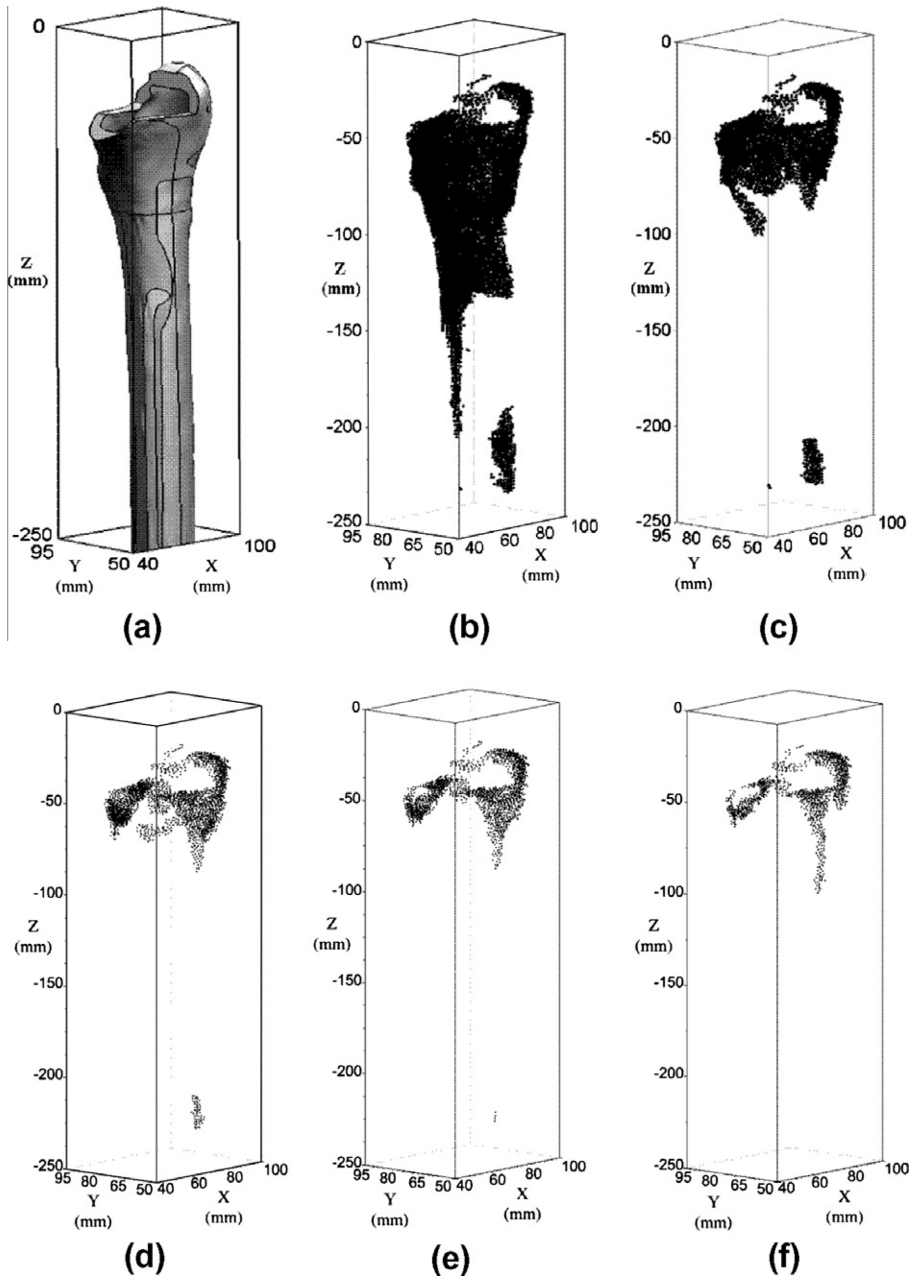

Fig. 6. 3D diagrams with dark dots representing the location with density loss in the cortical bone for different implant materials: (a) reference of the cortical bone; (b) cobalt–chrome; (c) solid titanium; (d) titanium foam with 20% porosity; (e) titanium foam with 40% porosity; (f) titanium foam with 60% porosity.

Fig. 6 clearly illustrates that the area with density loss significantly reduces in the cases of porous titanium. This study confirms that porous titanium could be a better solution for the hip replacement implant because it reduces the area with density loss, especially in the fixation area, which is crucial for the longevity of the implant.

4.2. Normalized bone density loss rate

In order to further evaluate the mechanical performance of porous titanium implants, we also tried to investigate its influence on bone remodeling rate, i.e., bone density loss rate here due to the femoral prosthesis. To accurately predict the bone density loss rate by applying Eq. (4), one needs to know the surface area density $a(p_b)$, which depends on the local porosity as indicated by Eq. (5). Generally speaking, a CT-scan is required to determine the local value of the surface area density in each finite element, which is challenging. Additionally, it is required to know the values of C_{inc} and C_{dec}, which should be obtained from clinical remodeling data. Moreover, the elastic modulus of the bone changes with the density of the bone. With bone remodeling, the elastic moduli of the cancellous and cortical bone evolute, which affect the strain field and in turn affect the bone density. To accurately quantify the bone density loss rate due to femoral prosthesis, a numerical iteration process is required to solve this mathematical coupling problem and to obtain the bone density loss rate as a function of time. The purpose of the current study is to evaluate the performance of titanium foam implant material by comparing with other existing materials. For this reason, we do not tempt to accurately quantify the bone density loss rate. Instead, we consider the following normalized bone density change rate:

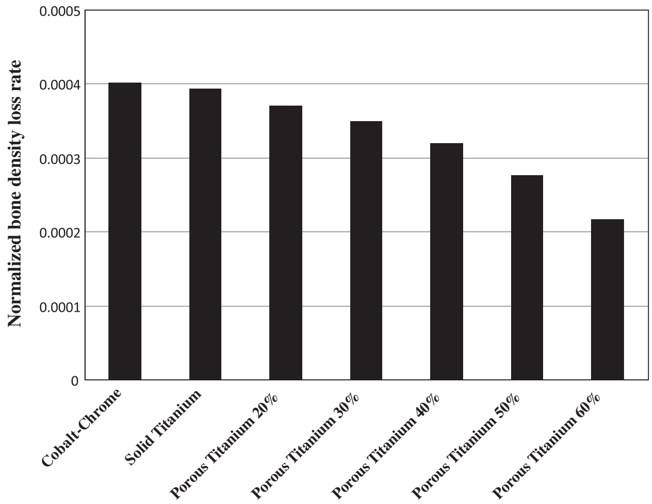

Fig. 7. The normalized bone density loss rate averaged in the cortical bone elements with density loss for different implant materials.

$$\frac{d\rho}{dt}\bigg/C_{inc}=\begin{cases}a[S-(1+s)S_{ref}] & \text{when } S>(1+s)S_{ref})\\ 0 & \text{when } (1-s)S_{ref}\leqslant S\leqslant(1+s)S_{ref}\\ a^{C_{dec}}/C_{inc}[S-(1-s)S_{ref}] & \text{when } S<(1-s)S_{ref}\end{cases} \quad (7)$$

The ratio of C_{dec}/C_{inc} is chosen as 3.5 following the study by Turner et al. [16]. To further simplify our comparison, a unit value of $a(\rho)$ is assumed for all the elements. The normalized bone density loss rate, the negative of $\frac{d\rho}{dt}/C_{Inc}$, varies from element to element. The averaged value of this variable from all the elements with density loss in the case of titanium foam with a 60% porosity is considered for the comparison. The reason to choose these elements is that most of these elements have density loss in all the other cases, as shown in Fig. 6. From the error analysis point of view, averaged value can eliminate the influence of finite element mesh on local strain estimation.

The calculated averaged bone density loss rate in the specified elements corresponds to the initial moment with the elastic moduli of the bones listed in Table 1. Fig. 7 shows the results for different implant materials. Cobalt–chrome implant leads to the highest bone density loss rate and the porous titanium with 60% porosity leads to the lowest rate. All the results for porous titanium are better than those for the solid titanium or the cobalt–chrome. Fig. 7 indicates that porous titanium implant can reduce the bone density loss rate and this reduction improves with the increase of the porosity of this material. These results also confirm that a porous titanium could reduce bone loss around the implant after a total hip replacement.

The effect of implant material on the bone loss rate is attributed to its elastic modulus. Fig. 8 shows the evolution of the bone loss rate as a function of the elastic modulus. The bone loss rate decreases with the reduction of the elastic modulus. It drops sharply when the modulus value is closing to that of the natural cortical bone (E = 17 GPa). The titanium foam with 60% porosity is the material with the lowest bone loss rate, whose elastic modulus is 17.6 GPa.

4.3. Porous titanium for the use of femoral implant

All the predicted results show that bone remodeling is affected by the elastic modulus of the implant material. As porous titanium has a lower elastic modulus, it has the higher potential to be applied as femoral implant material. Eq. (2) indicates that the elastic modulus of a porous titanium depends on its porosity. From the

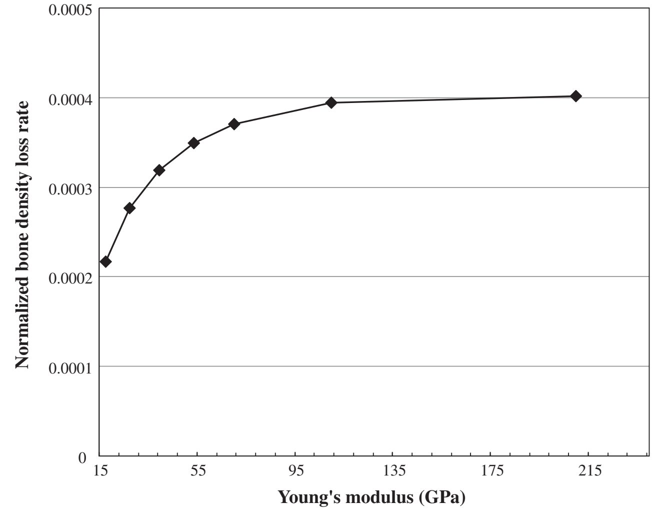

Fig. 8. The normalized bone density loss rate averaged in the cortical bone elements with density loss as a function of the elastic modulus of the implant material.

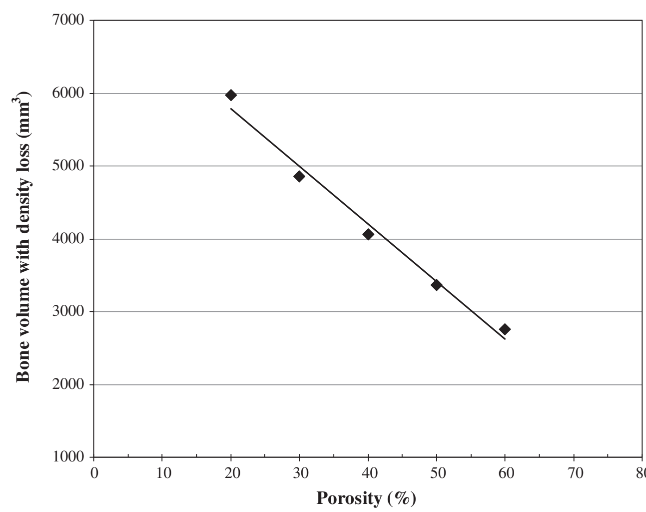

Fig. 9. The volume of cortical bone with density loss as a function of the porosity of the implant porous titanium.

material design point of view, it is important to know the influence of its porosity on bone remodeling directly. Here, the relationship between the total volume of the cortical bone with density loss and the porosity of the porous titanium implant is examined and the result is shown in Fig. 9. As we can see, volume of the cortical bone with density loss decreases with the increase of the porosity in the studied range of 20–60%. For the bone ingrowth, it is known that a porosity under 40% for the porous titanium is not ideal, because pores are usually not interconnected, therefore bones cannot grow into the implant [26]. The relationship in Fig. 9 can be approximately described by the following linear function:

$$V=-79.03P_t+7365.5 \quad (8)$$

Such a relationship enables people to extract the volume of the loss of density area for a porosity value, which could be very useful during the development of a new porous titanium-based prosthesis.

Increasing porosity will decrease the elastic modulus and therefore mitigate the problem of bone density loss. However, increasing porosity will also reduce the strength of the implant material as indicated by Eq. (3). To examine this issue, the maximum von Mises stress in the major part of the implant, which contacts with

Table 2
Maximum von Mises stresses and estimated yield strengths of titanium foams with different porosities.

Material	Maximum von Mises stress (MPa)	Estimated yield strength (MPa)
Titanium foam with 20% porosity	120	140–208
Titanium foam with 30% porosity	108	114–170
Titanium foam with 40% porosity	93	90–135
Titanium foam with 50% porosity	77	68–102
Titanium foam with 60% porosity	59	49–73

the cancellous bone, was obtained for porous titanium with different porosities from our FE calculations. The results are listed in Table 2. For commercially pure titanium grade 4, the yield strength of the fully solid material can reach to 650 MPa [27] and for solid titanium alloys, such as Ti–6Al–4V grade 5, its compressive yield strength can be 970 MPa [28]. Applying Eq. (3) and using the range of 650–970 MPa, the yield strength can be predicted for the foams with different porosities, see Table 2. According to Table 2, pure titanium foams with the porosity over 40% may not be acceptable as their strength could be lower than the maximum von Mises stress. The implant made from such a porous titanium might fail. However, titanium alloy foams with the porosity over 60% might still meet the strength requirement. Practically, experimental test is required to obtain the yield strength of the designed foams because predictions from Eq. (3) can only used as references. For example, the compressive yield strength from Imwinkelried's experimental tests for a pure titanium foam with the porosity of 50% is well above 100 MPa [27].

5. Conclusion

The purpose of this study was to find out if porous titanium could be a suitable implant material to reduce stress shielding after a total hip replacement. The finite element method was utilized to analyze the stress and strain state of femurs with an implant. A large range of porosity was considered to investigate the influence of the porosity on the bone remodeling. The majoring findings of this study can be summarized as follows:

- There is a sharp drop of the volume of the bone with density loss if a solid cobalt–chrome implant is replaced by a porous titanium implant.
- The percentage of the total volume with density loss in the cancellous bone is very low with porous titanium implant, which is less than 0.4% in the studied cases.
- Bone loss around the implant is clearly linked to the value of the elastic modulus of the implant. The higher the elastic modulus is, the more the bone will lose density.
- In the range of the studied porosity (20–60%), the bone density loss rate decreases almost linearly with the increase in the porosity.
- Increasing porosity will reduce the strength of porous titanium. With regard to material design for porous titanium-based femoral prosthesis, stress analysis is required to meet the strength requirement.

According to this study, porous titanium is a suitable implant material to reduce stress shielding after a total hip replacement. From the mechanical properties point of view, further study on fracture and fatigue of porous titanium is needed to confirm the potential of the porous titanium as a femoral implant [29,30].

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References

[1] Brown AS. Hip new world. Mech Eng 2006;128:28–32.
[2] Whiteside E. The effect of stem fit on bone hypertrophy and pain relief in cementless total hip arthroplasty. Clin Orthop 1989;247:138–47.
[3] Duda GN, Heller M, Albinger J, Schulz O, Schneider E, Claes L. Influence of muscle forces on femoral strain distribution. J Biomech 1998;31:841–6.
[4] Taylor M, Tanner KE, Freeman MAR, Yettram AL. Cancellous bone stresses surrounding the femoral component of a hip prosthesis: an elastic-plastic finite element analysis. Med Eng Phys 1995;17:544–50.
[5] Simões JA, Marques AT. Design of a composite hip femoral prosthesis. Mater Des 2005;26:391–401.
[6] Wen CE, Yamada Y, Shimojima K, Chino Y, Hosokawa T, Mabuchi M. Novel titanium foam for bone tissue engineering. J Mater Res 2002;17:2633–9.
[7] Spoerke ED, Murray NG, Li H, Brinson LC, Dunand DC, Stupp SI. A bioactive titanium foam scaffold for bone repair. Acta Biomater 2005;1:523–33.
[8] Takemoto M, Fujibayashi S, Neo M, Suzuki J, Kokubo T, Nakamura T. Mechanical properties and osteoconductivity of porous bioactive titanium. Biomaterials 2005;26:6014–23.
[9] Thelen S, Barthelat F, Brinson LC. Mechanics considerations for microporous titanium as an orthopedic implant material. J Biomed Mater Res A 2004;69:601–10.
[10] Bennett D, Goswami T. Finite element analysis of hip stem designs. Mater Des 2008;29:45–60.
[11] Sabatini AL, Goswami T. Hip implants VII: finite element analysis and optimization of cross-sections. Mater Des 2008;29:1438–46.
[12] Viceconti M, Ansaloni M, Baleani M, Toni A. The muscle standardized femur: a step forward in the replication of numerical studies in biomechanics. P I Mech Eng H 2003;217:105–10.
[13] Ziylan T, Murshid KA. An analysis of anatolian human femur anthropometry. Turk J Med Sci 2002;32:231–5.
[14] <http://www.zimmer.co.nz>. [accessed April 2010].
[15] <http://www.biomet.co.uk>. [accessed April 2010].
[16] Turner AWL, Gillies RM, Sekel R, Bruce W, Walsh WR. Computational bone remodelling simulations and comparisons with DEXA results. J Orthop Res 2005;23:705–12.
[17] Rolmann A, Moessner U, Bergmann G, Koelbel R. Finite-element-analysis and experimental investigation of stress in a femur. J Biomed Eng – T ASME 1982;4:241–6.
[18] Gibson LJ, Ashby MF. Cellular solids, structure & properties. Pergamon Press; 1988.
[19] Kashef S, Yan W, Lin J, Hodgson PD. Mechanical properties of titanium form for biomedical applications. Int J Mod Phys B 2008;22:6155–60.
[20] Ramos A, Fonseca F, Simões JA. Simulation of physiological loading in total hip replacements. J Biomed Eng – T ASME 2006;128:579–87.
[21] Huiskes R, Weinans H, van Rietbergen B. The relationship between stress shielding and bone resorption around total hip stems and the effects of flexible materials. Clin Orthop 1992;274:124–34.
[22] Martin RB. Porosity and specific surface of bone. Crit Rev Biomed Eng 1984;10:179–222.
[23] Robinson RP, Desyine GR, Green TM. Uncemented total hip arthroplasty using the CLS stem: a titanium alloy implant with a corundum blast finish. J Arthroplasty 1996;11:286–92.
[24] Scannell PT, Prendergast PJ. Cortical and interfacial bone changes around a non-cemented hip implant: simulations using a combined strain/damage remodelling algorithm. Med Eng Phys 2009;31:477–88.
[25] Weinans H, Huiskes R, Grootenboer HJ. Effects of material properties of femoral hip components on bone remodelling. J Orthop Res 1992;10:845–53.
[26] Shen H, Brinson LC. Finite element modeling of porous titanium. Int J Solids Struct 2007;44:320–35.
[27] Imwinkelried T. Mechanical properties of open-pore titanium foam. J Biomed Mater Res A 2007;81:964–70.
[28] <http://asm.matweb.com/search/SpecificMaterial.asp?bassnum=MTP641>. [accessed April 2010].
[29] Kashef S, Asgari A, Hilditch TB, Yan W, Goeld VK, Hodgsona PD. Fracture toughness of titanium foams for medical applications. Mater Sci Eng A 2010;527:7689–93.
[30] Kashef S, Asgari A, Hilditch TB, Yan W, Goeld VK, Hodgsona PD. Fatigue crack growth behavior of titanium foams for medical applications. Mater Sci Eng A 2011;528:1602–7.