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Influence of design features of tibial stems in total knee arthroplasty on tibial bone remodeling behaviors

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ABSTRACT

In total knee arthroplasty, the optimal length and material of tibial stem remain controversial. This study aimed to evaluate influences of lengths and materials of cementless stems on tibial remodeling behaviors. Three groups of lengths were investigated (i.e., 110, 60, and 30 mm), and four materials (i.e., titanium, flexible 'iso-elastic' material, and two functionally graded materials [FGMs]) were selected for each group. FGM is a kind of material whose composition gradually varies in space. In this study, the compositions of two FGMs were Ti and hydroxyapatite (FGM I), and Ti and bioglass (FGM II), respectively. Tibial models were incorporated with finite element analysis to simulate bone remodeling. Distributions of bone mineral density, von Mises stress, and interface shear stress were obtained. For the length, the long stem produced more serious stress shielding and stress concentration than the short stem, but it could provide better mechanical stability. For the material, FGM I could reduce stress shielding and stress concentration and reduce the risk of loosening. Compared with the length, the material had a pronounced effect on remodeling. This study provided theoretical basis for optimal design of stem to improve service life of tibial components and to reduce pain of patients.

1. Introduction

Total knee arthroplasty (TKA), which is an effective treatment of advanced arthritis of the knee, is expected with a consecutive distinct increasing applications [1,2]. However, 13% and 30% of TKA patients experience pain in short term and long term respectively [3], which may be caused by multifactorial etiology [4]. In addition, it was found that mechanical loosening was the most common reason for tibial component revision (24.6%) [5]. Several authors indicated that inadequate bone stock caused by stress shielding and stress concentration increases the risk of prosthesis loosening and periprosthetic fracture, and also triggers pain [6–8].

Stress shielding occurs when an implant, whose material property is usually stiffer than that of a bone, carries a part of the load originally carried by the host bone after implantation [7,9]. Thus, this implant forms a "shield" against the mechanical stimulus for bone tissues and causes bone resorption, thereby eventually leading to the possibility of aseptic loosening [6,10]. The prosthesis may increase the local stress level in the host bone, and this phenomenon is commonly known as stress concentration [9]. Studies showed that this phenomenon can stimulate the growth of the surrounding bone tissue, and can also be associated with pain and can increase the risk of periprosthetic fracture [7,8,11–13]. Therefore, the mechanical environment of the host bone must be improved by investigating the design of tibial components.

In recent years, studies regarding tibial components mainly focused on the materials [14,15], fixation [16,17], installation [18], and design of stems [7,19]. In the aspect of tibial tray materials, compared with the traditional titanium (Ti) and CoCrMo, functionally graded materials (FGMs) can reduce stress shielding of the host bone [14]. In the aspect of tibial component materials, a clinical study showed that in comparison with metal-backed tibial components, all-polyethylene counterparts significantly improved implant survival [15]. In the aspect of fixation ways, two kinds of techniques for primary TKA (i.e., cementless, and cemented fixations) were widely investigated [16,17]. In the aspect of final component alignment and tray position, the anatomic conflict between the tibial mechanical axis and intramedullary canal should not be ignored [18]. The optimal geometric length and material of stems, both affecting stress shielding and stress concentration directly, remain controversial particularly in the aspect of stem design [7,19]. Accordingly, the geometric length and the material of stems need further investigation and improvement.

Bone has functional adaptation, that is, bone adjusts its mass and architecture in response to the change of mechanical environment. Functional adaptation results in the change of bone material

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property, and then the bone adapts to the new mechanical environment [20,21]. Since the 1970s, some authors have quantified functional adaptation through bone remodeling governing equations, and these equations were improved by Huiskes et al. [13], Weinans et al. [22], and Mullender et al. [23], etc. "Lazy zone," nonlinear remodeling coefficient, and other concepts were also developed, which increased the accuracy of the bone remodeling algorithm in describing bone functional adaptation [24,25]. Given the functional adaptation, the investigations only examine the effects on bone tissue just only after the implantation provides an incomplete appreciation when describing the influence of prosthesis on the host bone [26]. However, a follow-up (long-term) quantitative simulation study can be implemented with bone remodeling governing equations [27–29], and many applications in the study of TKA have emerged. Nyman et al. [30,31] successively used this method to compare the effects of three kinds of fixation techniques (i.e., interlocking screws, cementless, and cement fixations) on bone loss and to assess the ability of bisphosphonates in reducing proximal bone loss in TKA. Chong et al. [32] focused on the influence of three methods of fixation (i.e., cement, cementless, and hybrid) on bone resorption, and they concluded that the cementless fixation with partial ingrowth or hybrid cementing fixation is preferred to maintain tibial bone stock. It was shown that the effects of tibial components on the host tibia can be simulated quantitatively and effectively with bone remodeling equation.

Accordingly, the present study aimed to evaluate the influence of design features (i.e., geometric length and materials) of tibial stems in TKA on tibial bone tissues, and to improve the mechanical environment of the host bone by seeking favorable material and length of stem.

2. Methods

The method incorporating finite element analysis (FEA) with quantitative bone remodeling algorithm was used in this study. The bone mineral density (BMD) distribution of proximal tibia was initially simulated, and then the three groups of cementless prostheses with different stem lengths, which included four types of materials in each group, were built to simulate the influences of different lengths and materials of the stem on tibial bone remodeling behaviors. The following three aspects were investigated in this study: BMD distribution of the proximal tibia before and after implantation, von Mises stress of the periprosthetic bone tissue, and shear stress on the interface between a stem and a bone. Fig. 1 illustrated the simulating steps.

2.1. Models

Due to the complexity of bone remodeling algorithm and the simplified structures used in previous studies [10,31,33], a simplified model of tibia was established in SolidWorks (Dassault Systèmes, SolidWorks Corp.) according to the morphology of tibial plateau and the coronal dimensions of tibia in computed tomography (CT) scan images (0.73×0.73 mm/pixel resolution, and 0.6 mm slice thickness). The CT data were from the left tibia of a male patient, who was 35 years old, 177 cm tall, and weighed 70 kg. The right tibia was fractured, and the CT scan data of the healthy left tibia were used. Considering that morphology of the tibial plateau was important for selection and design of the tibial component [34], and no differences in the anatomical shape of tibial plateau were found when comparing female and male or younger and older patients [35], as a result, using a young male tibial shape would not have a significant influence on the simulation results in comparison with using a tibial shape of an older osteoporotic patient. Then, the simplified model was imported into ANSYS (ANSYS Inc.) to mesh (Fig. 2a). Three groups of cementless tibial prostheses of the P.F.C Sigma Modular Knee System (DePuy International Inc.) in TKA were built in SolidWorks (Fig. 2b–d) directly according to the geometric parameters of tibial trays and stems listed in Table 1. And then the prosthetic models were imported into ANSYS. Overlap operation was performed on the tibia and on the three groups of prostheses in ANSYS. Thus, the three groups of tibial models containing prosthesis could be obtained.

Moreover, in consideration of the complicated contact conditions of osseointegration during bone remodeling process and hypotheses in previous studies [29,30,36], the interface between tibia and prosthesis was assumed to be tied and a perfect fit, simulating complete osseointegration [17,37]. Free meshing was conducted for the three groups of models, and the internal shape of prosthesis in each group should be considered to assign the material properties of the prosthesis before simulating the effects of prosthesis on the tibial BMD distributions. A 3D 10-node solid element (i.e., tetrahedron Solid 95) was selected for these models. This element type could simulate irregular shapes appropriately. It was also suitable for the models with plastic deformation or large strain [38]. To verify the FE models, mesh patterns with different element sizes were generated. The same material properties and the same loads were applied to the FE models with different mesh sizes. The material properties were $E = 2039$ MPa and $\nu = 0.3$, and the loads were applied according to Fang et al [39]. When variations of average strain energy density (SED) were less than 5% [39], the mesh was considered to be convergent. After convergence tests, the mesh size used in this study was 2 mm. Fig. 3 shows finite elements of the prosthesis and the tibia. The numbers of elements and nodes of the tibial models containing the prosthetic shapes are listed in Table 2.

The loads were applied according to Fang et al. [39] including joint contact force, shear force, medial collateral ligament force, and anterior cruciate ligament force acting on the proximal tibia with the magnitudes of 1233.3, 102.04, 6.4, and 139.3 N, respectively (Fig. 2e). These loads were the mean values of gait cycle loading. Of the joint contact force, 55% was applied on the medial side of the tibial plateau, and 45% was applied on the lateral side [42]. All the nodes at the bottom surface of the tibial models were fully constrained in all degrees of freedom.

2.2. Materials of the stem

In this study, three groups of stem lengths were investigated, and four kinds of materials (i.e., two homogeneous materials and two FGMs) were selected for comparisons in each group. The two homogeneous materials were Ti and the flexible 'iso-elastic' material, whereas the two different functionally graded composites were the mixture of Ti and bioactive hydroxyapatite/collagen (HAP/Col) (FGM I) and the mixture of Ti and bioglass (FGM II). The material properties of Ti were $E = 110$ GPa and $\nu = 0.27$ [43], whereas those of the flexible 'iso-elastic' material were $E = 14.22$ GPa and $\nu = 0.3$, which were close to those of cortical bone [10]. For the tibial trays, the material in all the groups was Ti.

The design of FGMs was based on the studies of Lin et al. [44] and Hedia et al. [9]. The material composition of the stem gradually transformed from Ti rich under the tibial tray to bioglass or HAP/Col rich toward the bottom of the stem, and the Young's modulus gradually decreased. The volume fraction distributed over z-direction (vertical distribution) is as follows:

$$V_t = (z/l)^m \tag{1}$$

$$V_c = 1 - V_t \tag{2}$$

where V_t and V_c were the fractions of Ti and HAP/Col (or bioglass), respectively; l was the total length of the stem; and m was the