Clinical Biomechanics 28 (2013) 514–518

Contents lists available at SciVerse ScienceDirect

Clinical Biomechanics

journal homepage: www.elsevier.com/locate/clinbiomech

Estimation of change of bone structures after total hip replacement using bone remodeling simulation

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ARTICLE INFO

Article history:
Received 8 June 2012
Accepted 2 April 2013

Keywords:
Bone remodeling
Total hip replacement
Loosening
Stress shielding
Image-based analysis

ABSTRACT

Background: The principal cause of femoral stem failure is the loosening of the total hip replacement due to bone resorption in the vicinity of the prosthesis (stress shielding). Bone rebuilds its structure continuously according to the daily mechanical stimuli. Therefore, surgical intervention alters the mechanical condition of bone severely. In this study, we propose a method to predict the change of bone structure after total hip replacement using bone remodeling simulation.

Method: The bone–stem complex structure model after total hip replacement was reconstructed based on CT-images used for preoperative planning by orthopedic experts. The bone remodeling simulation was conducted under the daily loading condition using our previous remodeling model, and the average equivalent stresses in the Gruen zone were evaluated.

Findings: The predicted bone loss relevant to stress shielding was consistent to follow-up clinical data. Moreover, the remodeling simulation when using the stems of different size for the same patient could detect the size-dependent change of stress in the Gruen zone. In particular, the zone under the neck of the stem showed significant changes of stress and large bone loss, accompanying the risk of loosening or fracture.

Interpretations: Prediction of bone structure changes after total hip replacement gives us significant information for longevity of prosthesis. Simulation results showed that the present computational framework could be considered to have potential in preoperative planning of total hip replacement.

1. Introduction

A total hip replacement (THR) has been increasing each year and the increase in the number of operations causes a corresponding increase in the number of failures (Gert et al., 2003; Marshall et al., 2004; Sakai et al., 2006). The mechanical failure of THR is mostly due to loosening, which disturbs stability of the stem and impairs its function. In THR surgery, femoral head is amputated partly and a stem is inserted into the canal of the femur; as a result, normal stress condition is no longer maintained, and consequently, the loss of bone density (stress shielding) may be induced. Most studies that quantify the change of bone mass after THR observed proximal atrophy to some degree dependent on stem design due to a lack of load in these regions (Braun et al., 2003; Engh et al., 2003; Herrera et al., 2004; Sinha et al., 2004). Therefore, to avoid stress shielding, mechanical compatibility of a stem with bone must be confirmed with great precision before THR surgery so that physiological load transfer from the prosthetic head to the femur may be achieved. The preoperative planning systems have been developed to confirm the fitness and stability of a prescribed stem in advance using individual images obtained by computed tomography (CT). In the long run, however, even an initial small disarrangement could have adverse effects on bone resorption–formation balance, leading to stress shielding and the disruption of stem stability. Bone tissues undergo habitual turnover cycles; normal bone structure is maintained by a balance of volumes of osteoclastic bone resorption and osteoblastic bone formation (Cowin et al., 1991; Frost, 2003; Hughes and Petit, 2010; Parafitt, 1994). Osteocytes play a role in the mechanical regulation of bone by receiving mechanical input signals and transmitting these stimuli to osteoclast/osteoblast (Cowin et al., 1991; Ehrlich and Lanyon, 2002). Many experimental studies have revealed mechanical factors influencing bone remodeling (Basso et al., 2005; Lang et al., 2004; LeBlanc et al., 1990; Shackelford et al., 2004; Vico et al., 2000). In addition, several simulation models considering the bone tissue responses to stress–strain stimuli (Carter et al., 1981; Huiskes et al., 1987, 2000; Ruimerman et al., 2005; Weinans et al., 1992) could reproduce some characteristics of trabecular bone structure or bone mineral density in human femur (Adachi et al., 1997). Under these circumstances, there have been several attempts for prediction of loosening after THR by bone remodeling simulation (Huiskes et al., 1992; Weinans et al., 1994). In these studies, however, the strain-dependent characteristics

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http://dx.doi.org/10.1016/j.clinbiomech.2013.04.003

of bone remodeling have not been fully incorporated. Recently, we developed a simulation model for bone remodeling based on mechanostat theory (Kwon et al., 2010a,b), which has been expanded from the earlier model (Adachi et al., 1997; Tsubota et al., 2009). Considering the effects of osteocyte apoptosis on bone resorption and targeted remodeling on bone formation, this model allows the acceleration of bone resorption and formation in response to the strain decrease and increase below and beyond the physiological strain range. Using this model, we could predict the trabecular structure for human femur successfully, which was highly-aligned to the direction of and magnitude-dependent on loads exerted on the femur.

Thus, the aim of this study was to predict changes of bone structure after THR by bone remodeling simulation based on our model. The bone–stem structure was constructed according to patients' femur and stem geometry obtained from a preoperative planning system based on CT images. The preoperative planning, femur shape, stem size and position were decided by orthopedic experts. The simulation results were validated in reference to clinical follow-up data for each Gruen zone.

2. Methods

2.1. Bone remodeling model

Trabecular structure was discretized using voxel elements, as shown in Fig. 1. Trabecular resorption/formation was expressed by removing/adding voxel elements and whether resorption or formation occurs was determined according to the equivalent strain estimated by finite element (FE) analysis. The quantity of removing/adding trabecular elements is also determined by activation frequency according to the equivalent strain. In our bone remodeling model, three mechanical usage windows were considered by referring to the mechanostat theory (Frost, 2003; Hughes and Petit, 2010): disuse window (DW), overuse window (OW), and physiological window (PW). We set the rules for bone resorption and formation in each window as follows (Fig. 2(b) and (c)).

In PW (equivalent strain range of ε_{pl}–ε_{pu}), the non-uniformity of the stress/strain distribution was taken as the driving stimulus (Adachi et al., 1997; Tsubota et al., 2009). For the voxel representing a point c on the bone surface, the degree of local stress non-uniformity Γ_c was quantified as

$$\Gamma_c = \ln\left(\sigma_c \sum_i^N w(l_i^c) / \sum_i^N w(l_i^c)\sigma_i\right) \tag{1}$$

where σ_c denotes the stress at the point c and σ_i denotes the representative stress at a point i with the distance l_{ic} apart from c (Fig. 1). The weight function $w(l_{ic})$ $[w(l_{ic}) > 0$ $(0 \le l_{ic} < l_L)]$ decays with distance, within the sensing distance l_L. The probability of bone resorption or formation at the point c is determined by $f(\Gamma_c)$, which is illustrated in Fig. 2(a), where Γ_u and Γ_l denote the lower and upper thresholds, respectively. When $f(\Gamma_c) = 1$, bone formation occurs, resulting in the addition of voxel elements around the surface point c, and when $f(\Gamma_c) = -1$, bone resorption occurs, resulting in the subtraction of voxel c. No change occurs at and around c when $f(\Gamma_c) = 0$. Moreover, in the cases of $0 < \Gamma_c \le \Gamma_u$ and $\Gamma_l \le \Gamma_c < 0$, bone formation and resorption occur stochastically according to probability $|f(\Gamma_c)|$, respectively. In this window, the activation frequency F_f or F_r, defined as a relative rate for bone formation or resorption, is set to unity, i.e., $F_f = F_r = 1$.

In DW (equivalent strain range of $<\varepsilon_{du}$), we assumed that only bone resorption occurs, and that the activation frequency of resorption increases with the decrease of strain at point c (ε_c) due to osteocyte apoptosis (Gu et al., 2005; Li et al., 2005; Noble et al., 2003). Thus, the probability of bone resorption is unity, i.e., $P_{DW} = 1$ and $f(\Gamma_c) = -1$. The activation frequency for bone resorption F_r increases exponentially with the decrease of ε_c and reaches F_{rmax} when $\varepsilon_c = \varepsilon_{min}$ (Fig. 2(b)).

In OW (equivalent strain range of $>\varepsilon_{ol}$), bone formation via targeted remodeling (Burr, 2002; Da Costa Gomez et al., 2005) was assumed to occur. That is, the probability of bone formation was set to unity, i.e., $P_{OW} = 1$ and $f(\Gamma_c) = 1$. The activation frequency for bone formation F_f, described by a sigmoid function and defined over the strain range of ε_{ol}–ε_{max}, was assumed to increase with ε_c and reaches F_{fmax} when $\varepsilon_c = \varepsilon_{max}$ (Fig. 2(b)).

These rules of bone remodeling in DW, PW, and OW were integrated by introducing transition regions (Fig. 2(c)). In the region of DW–PW ($\varepsilon_{du} < \varepsilon_c < \varepsilon_{pl}$) or the region of PW–OW ($\varepsilon_{pu} < \varepsilon_c < \varepsilon_{ol}$), the probabilities of bone resorption or formation were given by a linearly weighed sum. The details of mathematical expression of this model are found elsewhere (Kwon et al., 2010a). The resorption and formulation probabilities and the threshold values used for the present remodeling simulation are summarized in S-Table 1.

2.2. Image-based analyses

Patients' preoperative digital images of the femur were obtained by CT. Preoperative planning for THR was carried out by orthopedic experts based on the CT images, and the stem size and positions were determined. The latter were decided by THR surgery planning system (Otomaru et al., 2008). The stems used in this study were Centpillar GB HA Stems (Stryker Orthopaedics, Mahwah, NJ, USA) of size 6 to 8. The CT images were 0.7 mm thick and 0.7 mm × 0.7 mm pixel resolution. The FE models were constructed by voxel elements that were segmented into the femoral surface and femoral canal using the full CT information. Threshold values used to segment the femoral surface and femoral canal were 200 Hounsfield units (HU) and 800 HU, respectively. Image-based voxel element models are useful in explaining the effects of the mechanical environment on the trabecular structural changes that emerge from adaptive remodeling (Jang and Kim, 2008; Tsubota et al., 2009). The number of elements is approximately 200,000 each for the femur and stem. The bone material was assumed to be isotropic elastic medium. Poisson's ratio and Young's modulus were set to 0.3 and 20 GPa in the bone and 0.4 and

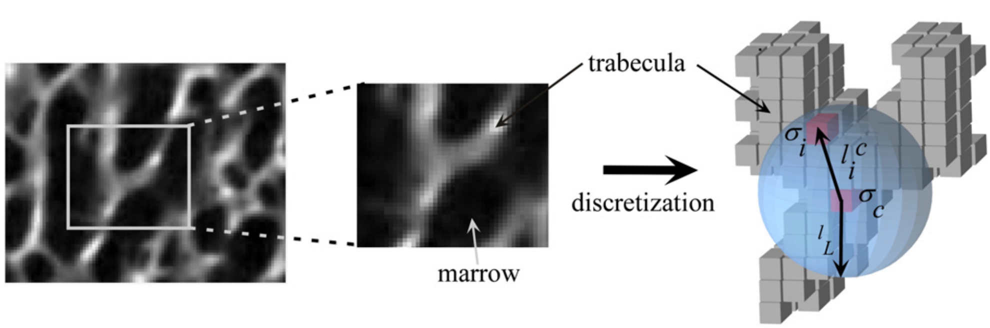

Fig. 1. CT image-based trabecular bone elements; bone elements were discretized by voxel element for FE simulation.