Computational Evaluation of the Effects of Bone Ingrowth on Bone Resorptive Remodeling after a Cementless Total Hip Arthroplasty∗

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In this study, we simulated a wide cortex separation from a cementless hip prosthesis using the bone resorption remodeling method that is based on the generation of high compressive stress around the distal cortical bone. Thereafter, we estimated the effect on late migration quantities of the hip prosthesis produced by the interface state arising from bone ingrowth. This was accomplished using cortical bone remodeling over a long period of time. Two-dimensional natural hip and implanted hip FEM models were constructed with each of the following interface statements between the bone and prosthesis: (1) non-fixation, (2) proximal 1/3, (3) proximal 2/3 and (4) full-fixation. The fixation interfaces in the fully and partially porous coated regions were rigidly fixed by bony ingrowth. The non-fixation model was constructed as a critical situation, with the fibrous or bony tissue not integrated at all into the implant surface. The daily load history was generated using the three loading cases of a one-legged stance as well as abduction and adduction motions. With the natural hip and one-legged stance, the peak compressive principal stresses were found to be under the criteria value for causing bone resorption, while no implant movement occurred. The migration magnitude of the stem of the proximal 1/3 fixation model with adduction motion was much higher, reaching 6%, 11% and 21% greater than those of the non-fixation, proximal 2/3 fixation and all-fixation models, respectively. The full-fixation model showed the lowest compressive principal stress and implant movement. Thus, we concluded that the late loosening and subsequent movement of the stem in the long term could be estimated with the cortical bone remodeling method based on a high compressive stress at the bone-implant interface. The change caused at the bone-prosthesis interface by bony or fibrous tissue ingrowth constituted the major factor in determining the extent of cortical bone resorption occurring with clinical loosening and subsequent implant movement.

Key Words: Cementless Hip Stem, Bone Remodeling Simulation, Bone Ingrowth, Migration

1. Introduction

Bone resorption around a cementless prosthesis stem with late clinical loosening and subsequent movement within the femur still remains an important problem in long term treatment following total hip arthroplasty (THA)1−5. Biomechanical research has verified that the inserted prosthesis changes the natural stress and load
transfer patterns in the cortical and cancellous bones, which are then absorbed and adapt to the changed stress patterns around the implanted prosthesis (6)–(9). With hip implants, the ‘stress shielding’ in the proximal region is generally known to contribute to femoral bone resorption.

Even though this bone resorption caused by the ‘stress shielding’ in the proximal femoral region is an important reason to play an important role in determining the stable fixation, the long-term loosening and migration of the hip stem caused by endosteal cortical bone resorption at the interface between the bone and prosthesis are more directly associated with the long-term successful fixation (1), (2). In particular, on the cementless implants, the bypassing load from the hip stem to bone and the concentrated high compressive stress around the bone-implant interface at the distal region have also been described as playing major roles in bone resorption after a primary cementless THA (10)–(13). Tanzer et al. (1) presented a 2–3 year follow-up study on the progression of femoral cortical osteolysis with porous-coated hip implants, concluding that the cortical bone resorption and subsequent stem migration occurred around the distal portion of the cementless prosthetic stem.

It has considered that the surface coating on a cementless hip prosthesis is one of the principal clinical design parameters necessary to reduce long-term loosening and the subsequent migration caused by enhanced bony ingrowth. Emerson et al. (3) indicated that a cementless hip stem developed radiolucent lines around the smooth tip of less than half the stem. Figure 1 shows a typical example of anteroposterior radiography of extensive cortical bone resorption around a cementless, porous-coated femoral prosthesis without cement. However, Engh and Bobyn et al. (14)–(16) reported that the porous coating of a cementless hip prosthesis could increase the stability of hip implantation after THA, with resorption occurring in only 18% of 411 cases studied. Weinans et al. (17) and Huiskes et al. (18) simulated the bone atrophy (bone resorption by stress shielding) phenomenon in proximal regions with different kinds of bonding characteristics. They concluded that the partial coating of cementless hip implants can significantly reduce bone atrophy relative to fully coated stems. However, some clinical follow-up data subsequently revealed a slightly different result. Engh et al. (15), (16) investigated the effects of coating in a 2–10 year clinical follow-up radiographic study on fully coated hip stems. They reported that bone ingrowth actually increased the survivorship rate and that the fully coated stem showed less bone resorption than the proximally coated ones.

In our previous report, we proposed a bone remodeling method for simulating the migration of a cementless hip stem based on the highly concentrated compressive stress caused by bone resorption occurring around the distal regions. This method involved various biomechanical design parameters, including those pertaining to material property and stem length (19). In this report, we assumed that the interface between the bone and hip stem was rigidly fixed by bony ingrowth. The purpose of the present study was to evaluate the long-term effects of bone ingrowth on more realistic interface conditions, with daily activity loadings being applied.

2. Methods

2.1 Bone remodeling simulation

In this study, we hypothesized that when the relative compressive stress on the bone was higher than a certain criteria threshold, cortical bone resorption would occur at the interface between the bone and prosthesis. The progress was evaluated at various local points of the cortical bone using a finite element analysis (FEA), and then the local compressive principal stress values ($S_{\text{p}}$) were compared to the criteria threshold of compressive principal stress ($S_{\text{c}}$) to determine the reduction ratio ($R_i$) of the cortical bone. In this study, the ‘geometry of cortical bones’ around implants was changed to simulate the phenomenon of cortical bone resorption. Thereafter, the reduction ratio ($R_i$) was used for shrinking.
In this study, the criteria threshold value of compressive principal stress ($S_c$) was determined to be $-50$ MPa on the basis of our FEA results and some researches\textsuperscript{(19)–(21)}. The lazy zone was determined by the $t$ parameter for the remodeling process, and $C_a$ and $C_b$ were proportional constants for the remodeling velocity (Fig. 3). The constant values of $C_a$ and $C_b$ were used as the assistant parameters to control the migration magnitudes of the hip stem. The constant values of 1.1, 0.5 and 1.2 were assigned to each $t$, $C_a$ and $C_b$ in this remodeling simulation. The simulation process was repeated until it was destabilized, following which stem migration would occur at the maximum five steps. These steps corresponded to periods of about 6 years on the basis of clinical results that reported migration magnitudes of about 1.3 mm / 2 years on the tip of the hip stem.

### 2.2 FEM analysis for bone remodeling

For the cortical bone remodeling around a cementless hip prosthesis, two-dimensional hip models, with and without hip prosthesis, were constructed with isotropic, elastic and four-node elements. All simulations of the two-dimensional hip FEM models were performed as plane stress problems with a thickness of 1 mm.

The straight cementless hip stem prosthesis was then implanted in accordance with a standard surgical method. Figure 4(a) illustrates the natural hip model, with implanted hip models having different interface conditions. In the bone remodeling simulation, the threshold value ($S_c$) controlling bone resorption was determined by the principal stress and strain analysis of the natural hip model.

For this simulation study, four types of hip implants within the bone were constructed with the following fixation statements at the bone-prosthesis interface: 1) non-fixation, 2) proximal 1/3, 3) proximal 2/3 and 4) full-fixation. These are shown in Fig. 4(b)–(e). It was assumed that the fixation interfaces in the fully and partially porous coated regions were rigidly fixed by bony ingrowth. The non-fixation model was constructed as a critical situation, with the fibrous or bony tissue not integrated at all into the implant surface. Non-linear gap elements with Coulomb friction ($\mu=0$) and the properties of a no-tension interface were assigned for the non-fixed interfaces in assuming a bony or fibrous tissue non-ingrowth\textsuperscript{(15)}.

The daily load history was generated using three loading cases, as illustrated in Fig. 5\textsuperscript{(21), (22)}. In considering the two-dimensional FEM models, the first load was the statement for a one-legged stance with a joint reaction force of 231.7 N directed at 24° from vertical, with a hip abductor force of 70.3 N at 28° from vertical. The other loading types were chosen at critical ranges of abduction and adduction motion with reduced force levels. The second loading case was constructed by a joint reaction force of 115.8 N directed at $-15^\circ$ from vertical with an associated

or expanding those elements with non-dimensional values, and was calculated to directly control the moving magnitude and direction of each node in the local coordinates of the bone; this is shown in Fig. 3. The changing direction of the cortical surface occurred along its inner normal direction of the cortical surface. When the compressive principal stress ($S_i$) was deemed over the criteria threshold value ($S_c$) in the local cortical bone, bone resorption was presumed to occur around the implant. If the compressive principal stress ($S_i$) was below the critical threshold ($S_c$), then the shape of the cortical bone remained unchanged.
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Fig. 4 The 2-D finite element models for a natural femoral hip (a) and implanted femoral hips with different interface statements (b)–(e) for the cortical bone remodeling (●: evaluated region of stress and strain, ■ and □: fixed interface region)

Fig. 5 The applied joint reaction and abduction muscle forces on the three loading types: (a) one-legged stance, (b) abduction and (c) adduction motions (G.S. Beaupré et al., 1990)

Table 1 Mechanical properties assigned to FEM models

<table>
<thead>
<tr>
<th>Material Type</th>
<th>Young’s Modulus (MPa)</th>
<th>Poisson’s Ratio (ν)</th>
<th>Static Strength (MPa) *</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cancellous bone</td>
<td>300</td>
<td>0.3</td>
<td>3 – 12</td>
</tr>
<tr>
<td>Cortical bone</td>
<td>17,000</td>
<td>0.3</td>
<td>150 – 200</td>
</tr>
<tr>
<td>Co-Cr alloy</td>
<td>220,000</td>
<td>0.3</td>
<td>827</td>
</tr>
</tbody>
</table>

* The static strength is under compression

abductor force of 55.1 N and −8° from vertical. The third loading case had a joint reaction force of 154.8 N directed at 56° from vertical with an abductor force of 46.8 N at 35° from vertical. The most distal nodes in the diaphysis were fixed in all directions for boundary conditions. In this simulation, the force values were carefully reduced in consideration of the thickness of the femoral shaft, as it was clearly not enough to cover the diameter of the femoral shaft as a three-dimensional problem.

In Table 1, the cortical and cancellous bones were assigned elastic modulus of 17 GPa and 300 MPa, respectively. A Poisson’s ratio of 0.3 was used for both the cortical and the cancellous bones. The thickness of the cortical bone was fixed at 1.5 mm around the femoral head, and the implanted straight hip prosthesis was simulated with a width of 15 mm and a femoral neck angle of 135°. All stress analyses were carried out with COSMOS/M ver. 2.8 (SRAC, USA) under the same loading quantity and boundary conditions. The computer program used to evaluate the geometric changes of the cortical bone caused by a change in the reduction ratio (Ri) was developed by FORTRAN and combined with the FEM codes.

3. Results

3.1 Stress and strain analysis

All compressive principal stresses and strains were computed with the 1) natural femoral hip model, 2) implanted hip models conducted immediately post-operative and 3) implanted hip models conducted after processing bone remodeling. The latter processing involved three loading types: 1) a one-legged stance as well as 2) abduction and 3) adduction motion. These relative stress and strain results were used in determining the criteria on compressive stress in governing bone remodeling and for simulating the long-term process of periprosthesi bone remodeling after a cementless hip prosthesis.

3.1.1 Natural femur hip In the natural femur hip model without prosthesis, the highest compressive principal stress of −45.9 MPa (−3.799 µ strain) occurred in the proximal 3/3 region of the lateral cortical bone under adduction motion. It was noted that all principal stresses, including the peak compressive principal stress, were under −50 MPa, which was reported as the criteria value of bone resorption or formation in various litera-
The compressive principal stress distributions of proximal 1/3 fixation models under the one-legged (a), abduction (b) and adduction motions (c).

Fig. 6

The evaluated strain (a) and compressive stress (b) of the natural hip model without prosthesis under the three loading types.

Fig. 7

Comparison of the peak compressive stress before (a) and after (b) the cortical bone remodeling process with different interface statements and the three loading types.

Fig. 8

3.1.2 Femur with implanted prosthesis in the immediate postoperative period

With the cementless hip prosthesis, stress analyses showed that the peak compressive principal stress occurred in the lateral or medial distal cortical bone along the hip prosthesis, a typical pattern for stems inserted in a cementless mode (Fig. 7 (a)–(c)). In addition, peak compressive principal stresses with the added abduction motion were found along the medial cortical region, while those with the added adduction motion occurred in the lateral cortical regions. It was shown that those regions of high compressive principal stress were relatively more sensitive to changes in the abduction and adduction motion than in the one-legged stance.

The non-fixation model having the worst interface integration by bony non-ingrowth showed the highest compressive principal stress with all loading types. These were $-78.6$ MPa for the proximal 1/3 lateral cortical region of the non-fixation model as shown Fig. 8 (a). The compressive principal stresses of all implanted hip models were over $-50$ MPa with added abduction and adduction motions, except for the one-legged stance. The proximal 1/3 fixation model showed universally higher compressive principal stress levels in the proximal 2/3 and 3/3 of the cortical regions, even though the proximal 2/3 fixations model showed a slightly higher stress around only the proximal 3/3 regions. In the full-fixation model, the compressive principal stresses were approximately $-53$ MPa and $-55$ MPa on the abduction and adduction motions, respectively, which were slightly higher than the $-50$ MPa criteria governing bone resorption or deformation. It was therefore considered that the full-
fixation model, even though well fixed in the interface between bone and prosthesis, can also develop resorptive bone remodeling around the stem.

3.1.3 Implanted femur hip model after processing the bone remodeling

After cortical bone remodeling in the long term, there was respectively little change in the stress distribution and levels in the natural femoral hip model of all loading types as well as the one-legged stance mode. In the non-fixation model, the compressive principal stresses greatly increased to about 2.1 times, remaining entirely higher compressive principal stress significantly higher than those of other models under the abduction and adduction motions as shown in Fig. 8 (b). However, upon completion of the bone remodeling, the highest compressive principal stress of −174 MPa was found in the proximal 1/3 fixation model around the proximal 3/3 lateral regions under the adduction motion. In addition, the increased compressive stress, which was over the static strength of about −200 MPa (23), showed a possibility of bone fracture in the periosteal regions. The full-fixation model showed the lowest increase in compressive principal stress (about 1.6 times) amongst all loading types.

3.2 Stem migrations caused by cortical bone resorption

The magnitudes of prosthesis migration caused by cortical bone resorption were calculated in the tip of the hip stem in relation to various conditions involving: 1) the interface statement (non-fixation, proximal 1/3, 2/3 fixation and full-fixation) and 2) the loading types (one-legged stand, abduction and adduction). These magnitudes were compared to clinical follow-up data in the literature. A criteria value for compressive stress controlling cortical bone resorption was determined to be −50 MPa on the basis of our results in the stress and strain analyses.

3.2.1 Comparison of stem migration quantities in relation to the interface statement

Figure 9 shows an example of the cortical bone remodeling process in the proximal 1/3 fixation model under abduction and adduction motions. The bone resorption occurred along the medial and lateral cortical bones, before the stem moved into the disruption regions of the bone as shown in Fig. 9 (f) and (k). The peak magnitude of implant movement was found in the proximal 1/3 fixation model during adduction motion. However, the implant movements of the non-fixation model were all higher than those of the proximal 1/3 fixation model for all loading types. The migration magnitude of the stem on the proximal 1/3 fixation model in adduction motion was much higher, reaching 6%, 11% and 21% greater than that of the non-fixation, proximal 2/3 fixation and all-fixation models, respectively. This is shown in Fig. 10. Cortical bone resorption did not occur in the natural hip model because the principal stress levels in all regions were under −50 MPa. The full-fixation model showed the lowest implant movement in all loading modes. These finding indicated that the bony or fibrous ingrowth between the bone and implant could be important determinants in increasing long-term implant stability.

3.2.2 Respective stem migrations changed in loading types

The magnitudes of migration were relatively more sensitive to changes in interface statements than by loading types. In all models, the magnitudes of implant movement on the adduction motion were much higher than those on the abduction motion. Moreover, implant movements did not occur in the one-legged stance. On the proximal 1/3, 2/3 and full-fixation models, the implant movements were 21%, 18% and 11% lower, respectively, in abduction than in adduction motion (Fig. 10). However, the non-fixation model showed a universally higher implant movement as compared to other models and there was no significant difference (3%) in the migration between the abduction and adduction motions.

Fig. 9 One example of the cortical bone remodeling process on the proximal 1/3 fixation model under abduction (b)–(f) and adduction motions (g)–(k)

Fig. 10 Comparison of the predicted migration of hip implants on different interface statements under abduction and adduction motions
4. Discussion

In this study, we focused on simulating the loosening and subsequent migration of a cementless hip prosthesis, a wide cortex separation from hip prosthesis based on the generation of high compressive principal stress around the distal cortical bone. We also estimated the long-term migration quantities of the hip prosthesis using cortical bone remodeling FEM simulation methods with respect to interface statements and loading types.

The late loosening and subsequent movement of the hip prosthesis caused by bone remodeling occurring after the implantation of a cementless hip prosthesis remains an important problem that has been reported in many clinical follow-up studies. Engh and Bobyn et al. demonstrated that the resorptive bone remodeling changes were attributed to the late loosening and migration categorized them into degrees of severity. As the fourth degree, cortical bone resorption extended below the proximal 1/3 and 2/3 regions into the diaphysis, with the changes characteristic occurring in the cortices just above the level of the press fit, where the cortex was most widely separated from the straight stem. In addition, they confirmed through radiographic evidence that a cementless hip prosthesis with the appearance of bony or fibrous tissue ingrowth produces good clinical results. Emerson et al. found that the cementless stems had radiolucent lines exclusively around the smooth tip of the stem in less than half of the stem as well as having osteolysis exclusively in the proximal 1/3 zone, with none down the endosteal canal or around the stem tips. It was considered that the decrease of bone mineral density in the inner proximal 1/3 cortical bones may not be as significant of a cause of the clinical loosening and movement of stem as the high stress concentration and micro-motion caused by the changed load transfer at the bone-prosthesis interface, which has been reported in clinical and biomechanical studies.

There is a limitation to two-dimensional (2-D) analysis in this study because it is inherently less stiff than a three-dimensional (3-D) model containing the enclosed cortical construction. However, the periprosthetic stress distributions and levels predicted by our 2-D models should at least be qualitatively consistent with those obtained in a 2-D or 3-D analysis. Furthermore, in improving our remodeling simulation theory, we should construct three-dimensional FEM models and carefully consider the effects of torque on the bone remodeling process and implant stability for our next paper.

With respect to the bone remodeling criteria, it is difficult to determine exactly the absolute value of the physiological strain, or stress governing bone resorption or deformation, because the physiological responses of bones vary in accordance with different bone constructs and daily load bearing during daily activities. In order to simulate the bone resorptive remodeling, we applied the compressive principal stress of \(-50\) MPa (\(-3\,799\) \(\mu\) strain) on the base of our stress/strain analysis in the natural hip model (Fig. 7) and some references. Beaupré et al. simulated time-dependent bone remodeling in a natural hip model. They used a stress stimulus of 55 MPa/day (\(-400\) \(\mu\) strain /10000 cycles) for bone apposition and resorption. Sugiura et al. reported that cortical bone resorption occurred in concentrations of high compressive principal stress regions that were over the critical threshold value, about \(-3\,600\) \(\mu\) strain (\(-50\) MPa), shown with experimental testing and FEM analyses around mandibular fixation screws.

Several bone remodeling simulation methods based on local stress, or strain for bone remodeling, have been proposed in the literature in order to estimate the loosening and de-bonding phenomena of hip implants between the bone and prosthesis. Huiskes et al. hypothesized that when the interface stress between the bone and implant exceeded the strength of the bonding, interface disruption or debonding would occur locally. According to this theory, it was also assumed that if the relative motions caused by the disruption interface exceeded a certain threshold level, they would affect the osteoblastic and/or osteoclastic activities of the adjacent bone and provoke interface-bone resorption. However, they did not consider the geometry changes of bones caused by bone resorption at the interface between the bone and hip stem. It has generally been known that the bone structure can be adapted to the changed load or stress patterns. In our study, we considered changes in the bone geometry at the bone-implant interface in regards to bone resorption remodeling, determining that it should accelerate the long-term loosening and subsequent migration of the hip implant (Fig. 9).

It has been considered that surface coating of a cementless hip implant serves as one of the main parameters to stabilizing the total hip replacement system. Huiskes and Weinans et al. simulated the bone atrophy (bone resorption by ‘stress shielding’) phenomenon in the proximal femoral region after the implantation of a cementless hip stem. They concluded that the bonding characteristics induced by bony ingrowth were important parameters to determining long-term bone resorption in the proximal region. In their results, they concluded that the partial coating of cementless hip implants can significantly reduce bone atrophy relative to that of fully coated stems. Although the various coating statements affect bone resorption in the proximal region constructed by the cancellous bone, it was thought that excessive cyclic distal tip motions could produce peripheral endosteal cortical bone resorption at the bone/implant interface. The endosteal cortical bone resorption at the bone/implant interface can directly induce the loosening and subsequent migration of the hip stem. Taking this point of view, we focus on sim-
ulating the long-term migration of the hip stem caused by the endosteal cortical bone resorption remodeling around hip stem and evaluating the effect of bone ingrowth on the loosening and migration of the hip stem.

The present findings show that the concentrated high compressive principal stress dramatically led to the loosening and subsequence migration of the cementless hip prosthesis (Fig. 9). The peak compressive principal stress was found to be present along the distal cortical bone, and the migration quantities of the stem tip ranged from 0.2 to 4.4 mm. Although it was difficult to directly compare our findings to the clinical loosening data available in the literature due to the different types of hip prosthesis described and the shorter follow-up data, producing a bigger error range, our findings may be used for designing hip stems that prevent stem migration following a THA implantation. Tanzer et al.\(^1\) and Goetz et al.\(^2\) indicated that the existence of endosteal cortical erosion of the femur through clinical follow-up data on a primary cementless THA. Their results were very similar to our findings in this study.

With respect to the bone-implant interface statements, some studies have reported that a porous-coated hip replacement clearly increases long-term stability in both bone-ingrowth and fibrous tissue fixations\(^{14}–^{16}\). However, the influence of extended porous coating on the femoral bone resorption is still a controversy. In this study, the highest levels of compressive principal stress and bone resorption were observed in the proximal 1/3 fixation model, while the non-fixation model showed entirely high compressive principal stresses before/after the bone remodeling process. Even though all of the bone-prosthesis interfaces were well fixed by bony or fibrous tissues, there was still the possibility of interface disruption due to the high stress concentration caused by the critical motion occurring with added abduction or adduction. The results further showed that extent coating and stable walking can be important factors in enhancing the long-term stability of the cementless hip stem. Our results suggest that porous coating treatment to enhance bony or fibrous tissue ingrowth may well be used to control the magnitude of periprosthetic bone remodeling on a cementless hip prosthesis.

In conclusion, the FEA bone remodeling method based on the high compressive principal stress at the interface between the bone and cementless hip prosthesis can estimate the extent of loosening and the subsequence movement of the stem over the long term. The changes at the bone-prosthesis interface by bony or fibrous tissue ingrowth were the major factor in determining the cortical bone resorption occurring with the clinical loosening and subsequence implant movement. In addition, the load types caused by daily activity were identified to be important parameters, having an affect on the migration magnitudes as well as directions.

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References


