Long Stemmed Total Knee Arthroplasty With Interlocking Screws: A Computational Bone Adaptation Study

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Abstract
The ability of an interlocking screw fixation technique to minimize bone loss related to stress shielding in the tibia was investigated and compared to the abilities of cement and press-fit fixation. Full bony ingrowth has been associated with greater stress shielding than partial ingrowth; therefore, the effect of intimate bonding of the stem to bone on subsequent bone loss was also studied. A damage- and disuse-based remodeling theory was coupled with a two-dimensional finite element model of the tibia to predict changes in bone remodeling following long stemmed total knee arthroplasty (TKA) for four different fixation techniques (cement, press-fit, interlock with bony ingrowth, and interlock without bony ingrowth). Remodeling changes commenced with the model state variables—bone area fraction, mechanical stimulus, damage, and remodeling activity—at steady-state values predicted by the intact tibia simulation. After TKA and irrespective of fixation technique, the model predicted elevated remodeling due to disuse, in which more bone was removed than replenished. In regions below the tibial tray and along the cortices, the interlocking stem with full bony ingrowth and the cemented stem caused the least amount of bone loss. An interlocking stem with a smooth, matted finish did not reduce the bone loss associated with interlocking fixation.

Introduction
The number of total knee arthroplasty (TKA) surgeries has increased every year since 1980 [19]. Subsequently, the number of revisions has increased because mechanical loosening of primary prostheses has not been eliminated, despite improvement in the success rate of the surgery. At revision when bone stock is deficient, implanting a long medullary stem is often necessary in order to stabilize the prosthesis while allowing reconstruction of the bone defect (e.g., with a bone allograft) [13]. Unfortunately, long stems increase the incidence of stress shielding [24], though the design of the stem and the means of fixing the stem to bone could minimize this phenomenon.

Bone loss is closely associated with the aseptic failure mechanisms of TKA. Although this loss could be the result of debris-induced osteolysis, shielding of bone from physiologic stress by implant components is another important factor in causing the reduction in bone density and strength that leads to failure [22]. Unfortunately, consensus has not emerged from clinical followup studies on which fixation technique in revision TKA provides the lowest rate of loosening [3]. In the case of primary TKA with short stems (40–60 mm in length), X-ray absorptiometry studies that measured bone density beyond three years after surgery demonstrated the occurrence of bone loss [29,30]. Greater bone loss is expected to occur with long stems (>60 mm) based on a strain gage study [20] and a finite element (FE) analysis [2].

Total hip arthroplasty (THA) has been more widely studied, thereby providing much of what is known about stem fixation. Superior fixation is achieved when the stem is cemented to the contiguous bone [10]. However, difficulty in removing cemented components and the identification of three ‘weak-link zones’ (the cement–implant
interface, the cement–bone interface, and the cement mantle itself) as potential sites for the initiation of loosening [14] have encouraged the development of cementless alternatives. In the absence of cement, a ‘press-fit’ stem is required to achieve sufficient stability; however, an increase in stem-to-bone stiffness ratio has been positively correlated with stress shielding [37] and a greater incidence of thigh pain at the stem tip [36]. A reduction in material stiffness to compensate for the increase in diameter is not feasible because of the high shear stress that develops between isoelastic stems and bone as shown by a computational strain-adaptive bone remodeling study [18]. Similar problems in TKA have been identified with FE analyses [27,34].

A new procedure that stabilizes a smaller diameter stem with interlocking screws is being developed to address the concerns of stress shielding and pain (Fig. 1) [31]. Originally, the idea to interlock stems with screws in revision total joint arthroplasty arose from the desire to provide early weight bearing and immediate fixation, thus allowing time for ingrowth and restoration of the compromised bone. In addition, an increase in bone density with impaction grafting was achieved with interlocking stems [35]. The current motivation for interlocking screw fixation is its ability to secure a small diameter stem without using cement. While this method reduces the stem’s stiffness (by decreasing its moment of inertia) without lowering the material stiffness, little is known about the effect of interlocking screws on bone loss. Given that the interlocking screws provide sufficient stability, the stem does not necessarily require porous coating for bony ingrowth. Furthermore, several THA studies found that press-fit stems with full bony ingrowth caused greater shielding than partial ingrowth (i.e., proximal porous coating) [12,17]. Possibly, a nonbonded stem–bone interface would cause less bone loss than a bonded interface.

To quantify the level of tibial stress shielding as it relates to stem fixation, we developed a two-dimensional (2D) adaptive bone remodeling simulation that predicts remodeling changes following long stemmed TKA. Our primary research question was: will interlocking a long, small diameter stem with titanium screws reduce bone loss associated with stress shielding when compared to cement and press-fit techniques? In addition, we addressed the question of whether fixation via bony ingrowth into a porous coating minimizes or exacerbates bone loss. Lastly, changes in remodeling parameters after TKA were analyzed over time to understand the mechanism of bone loss.

Materials and methods

Finite element model of the tibia

Using PATRAN 8.5 (MSC, Santa Ana, CA), a FE model, consisting of 3884 quadrilateral and two triangular plane strain, linear elements was developed. The mesh modeled the following: a frontal outline of a tibia, which covered the proximal metaphysis to the distal third of the diaphysis and was acquired from digitizing an X-ray of a representative tibia; a tibial tray attached to a vertical stem with a length of 120 mm and a diameter of 12 or 16 mm; two titanium screws (5 mm in diameter) passing horizontally through the stem at 25 and 50 mm from stem tip; and a 2 mm cement layer below the tray and surrounding the 12 mm wide stem (Fig. 2). Stem loosening was not modeled. The condyles of the intact model had the same profile as the tibial insert for convenience, and
the titanium screws were in intimate contact with bone, thereby simulating rigid fixation to both the bone and the stem.

Isotropic material values (Modulus, E, and Poisson’s ratio, m) were used [21]: PMMA bone cement (2.150 GPa, 0.46), UHMWPE (2.3 GPa, 0.25), and titanium (79 GPa, 0.36). To account for the out-of-plane contribution to stiffness in the 2D model, a thickness value of 27 mm was assigned to the elements, and a bony side plate consisting of 2279 quadrilateral elements with graded thickness (5.5 mm distal to 1 mm proximal) and cortical bone properties [1] covered the model, as described in a THA study by Huiskes [17]. By using this method and applying a joint reaction force (JRF) equal to 3 times body weight for a 70 kg person, the FE analysis calculated medial tibial surface strains in the physiologic range, as measured by Burr et al. [7] at the midshaft.

Model of adaptive bone remodeling

A mechanistic model of damage- and disuse-based bone remodeling, the behavior of which is described by Hazelwood et al. [15], was incorporated into each element of the FE model (Fig. 3). The adaptive model assumes that basic multicellular units (BMUs), the organized teams of bone cells that remove then replenish bone, are activated to remove bone tissue that is insufficiently loaded (i.e., in relative disuse [11]) or to replace bone which is fatigue damaged during daily activity [28]. Experimentally, both insufficient loading [23] and fatigue microdamage [5,25] have been shown to stimulate the remodeling process. Calculations for the damage and disuse stimuli in the model are adapted from the mechanical stimulus approach of Carter et al. [8].

The strain state (S) of bone, which is defined as the maximum absolute value of the principal strains, and the number of cycles per day (R₁) at which the strain is loaded, summed over n discrete daily loading activities, combine to produce a loading stimulus,

\[ S \cdot R \phi \]

where the exponent q was set to 4 because this empirical weighting factor provides a reasonable sensitivity relationship between loading magnitude and loading frequency for changes in bone density [38]. The rate at which damage \( \frac{dD}{dt} \) forms in bone is assumed equal to \( k_D \phi \) where \( k_D \) is a damage coefficient that was determined under initial and equilibrium remodeling conditions to be 1.85 · 105 mm/mm2 [15]. Offsetting damage formation is the rate of damage removal \( \frac{dD}{dt} \) by BMU remodeling, which is a function of the amount of damage (D) present in the region of interest, the activation frequency of BMUs (Ac.f) in the region, the cross-sectional area resorbed by each BMU (Rs.Ar), and a specificity factor (Fs) that accounts for the spatial association of BMUs to microcracks. The difference in damage formation and removal rates is integrated to determine the current amount of existing damage. Thus, damage introduces bone loss because its removal introduces transient porosity due to an increase in resorption cavities and, in cortical bone, permanent porosity in the form of Haversian canals.

The “attractor state stress stimulus” of Beaupre et al., a function analogous to \( \phi \) in Eq. (1), enabled the use of experimental data to define combinations of cyclic strain level...
and frequency that maintain a constant bone mass [4]. Here, we use a similar approach: disuse is simulated when values of the loading stimulus fall below an equilibrium value ($\phi_0$). Based on the results of Beaupre et al., $\phi_0$ was set to $1.88 \cdot 10^{-10}$ cycles per day. When trabecular bone is in disuse, the bone apposition rate ($Q_F$) decreases as follows:

$$Q_F = \begin{cases} \frac{AF}{TF} & \text{for } \phi < \phi_0, \\ \text{constant} & \text{for all } \phi, \end{cases}$$

where $AF$ is the cross-sectional area of new bone within a completed BMU and $TF$ is the formation period. Otherwise, for $\phi \geq \phi_0$, $Q_F = \frac{AF}{TF}$. The BMU resorption rate is assumed constant whether bone is in a disuse state or not:

$$Q_R = \frac{AR}{TR}$$

where $AR$ is the cross-sectional area of a BMU resorption space and $TR$ is the resorption period. Thus, in disuse, elevated Ac.f with a diminished apposition rate (Eq. (2)) causes a negative bone balance (bone removal exceeds bone formation at each remodeling site).

The current rates of bone resorption and formation in a representative cross-section within each bone element are obtained from the history of BMU activation. Integration of this history over the respective phases of a BMU lifetime provides the numbers of resorbing ($N_R$) and refilling ($N_F$) BMUs. Synthesizing results from several histomorphometric studies, we used 25 and 64 days, respectively, for the resorption and refilling periods [15]. The difference in the rate of bone removal ($Q_R N_R$) and addition ($Q_F N_F$) determines the rate of change in the amount of bone per unit area or bone area fraction (BAF). Relating modulus to BAF (Eq. (i) in Fig. 3), the FE analysis also uses a $m$ of 0.03 to calculate the strain state for each bone element throughout the model of the tibia. In an iterative fashion over the desired time period and with values of model coefficients and Ac.f relationships for damage and disuse from Hazelwood et al. [15], the coupled remodeling and FE models simulate the remodeling response of bone to its mechanical environment.

**Intact tibia and TKA remodeling simulations**

With the distal end constrained, a JRF of 2058 N (3 times body weight [26]) was applied to the condyles in three different distributions to simulate the daily loading history. Cases were: (1) JRF evenly distributed normal to each condylar surface, (2) 70% and 30% of JRF distributed across the medial and lateral condyle, respectively, inclined 5.0° from the vertical such that the horizontal component of force was directed medially, and (3) a mirror image of (2), i.e. 30% medial, 70% lateral, with the horizontal component directed laterally. The loading frequency of each case was 3000, 500, and 500 cycles per day, respectively. These loading activities were chosen to capture peak strains during walking as well as strains from valgus and varus moments that occur during daily activity, such as stair climbing [39] and crossover cutting [6]. Furthermore, the tilt in vertical JRF accounts for the medial–lateral ground reaction force when the foot is planted [9], and the shift from medial to lateral bias in the JRF distribution provided the strain distributions that formed the cortices of the diaphysis of the tibia (Fig. 4(A)).
A user-defined material behavior subroutine (UMAT) integrated the bone remodeling algorithm into the FE analysis, and the simulations were run using ABAQUS 5.8 (HKS, Pawtucket, RI). To ensure enough elements were used to calculate the principal strains for the remodeling algorithm, a test model having 3.8 times more elements was run. The root mean square of the difference in nodal BAF values between the two models was <0.0025 at 400 days suggesting the study model had an adequate mesh density.

Starting with homogenous material properties for a BAF of 95.2 (representing the canal porosity of completed osteons), the bone in the intact tibia model remodeled until each state variable (e.g., BAF, D, U, Ac.f) changed less than 0.1% per year. Adjusting the UMAT so the implant elements were set to their fixed material properties and starting from this equilibrium point (Fig. 4(A)), simulations were run for 2000 days of remodeling to study four different TKA methods: (1) a cemented 12 mm diameter stem (cement); (2) a press-fitted 16 mm diameter stem (press-fit); and an interlocked 12 mm diameter stem with the stem either (3) intimately bonded to bone (lock bonded) or (4) in hard, frictionless contact with bone (lock contact). Except in the lock contact case, in which the stem and bone had two separate contact surfaces that were paired, no relative motion between the components (stem–bone, stem–cement, and cement–bone) was allowed. Since friction between the stem and bone could vary and is difficult to ascertain, frictionless, hard contact (i.e., shear or tensile stress was not transmitted across the interface) was chosen as an extreme for comparison to full bonding. The tibial tray was cemented to the sectioned metaphysis in all cases. Bone loss was quantified as percent decrease in BAF of the intact tibia.

Results

With the given loading conditions, the model produced the essential features of an intact tibia (cortical bone in the diaphysis along each side of a medullary canal and trabecular bone in the metaphysis, Fig. 4(A)) and simulated proximal bone loss after TKA (reduction in BAF and thinning of cortices at 2000 days, Fig. 4(B)). Long stemmed TKA caused bone loss in all regions examined, with trabecular bone suffering more loss than cortical (quantified at 2000 days, Fig. 5). Of the fixation techniques, the press-fit stem with a bonded stem–bone interface had the most bone loss, except in region 2 (trabecular bone in the distal metaphysis), where the interlocking stem with a non-bonded interface lost 8% more. The interlocking stem with a bonded interface had 1% more bone loss than the press-fit stem in region 2, and the loss of bone in the other regions (below tibial tray and at the stem tip), as well as two additional regions (distal trabecular and proximal cortical bone, Fig. 2(B)), was similar to that of the cement stem (0.6% and 0.1% more, respectively), which caused the least amount of bone loss. Of the two interlocking stems, the one simulating full bony ingrowth had 1.8% and 5.9% less trabecular bone loss in regions 1 and 2, respectively.

The pattern of change for the state variables over time following long stemmed TKA was similar for all fixation techniques. Initially, BAF in the model rapidly decreased, but after the first 6 months the rate of decrease was gradual (Fig. 6(A)). At 2000 days, region
2 had the greatest relative decrease in BAF. Immediately following TKA, \( \phi \) dropped below the equilibrium threshold in all regions (Fig. 6(B)), although after 1 month region 4 (cortical bone of the diaphysis near the stem tip) was no longer in a state of disuse and as a result experienced the least amount of bone loss. The drop in \( \phi \) caused damage to decrease in all regions because the rate of damage formation declined (Eq. (ii) in Fig. 3). In addition, the fact that disuse (i.e. stress shielding) stimulated \( \text{Ac.f} \), damage removal initially increased (Eq. (iii) in Fig. 3). Therefore, in the regions examined, damage had little influence on bone remodeling and bone loss immediately following TKA. However, as \( \phi \) increased with time and BMU activity associated with disuse declined, damage had a greater influence on bone remodeling.

**Discussion**

The aim of the present study was to determine if interlocking screw fixation minimizes bone loss associated with stress shielding in revision TKA. In doing so, we performed a parametric analysis comparing interlocking fixation to press-fit and cement fixation. Stress shielding in joint arthroplasty is a phenomenon in which bone experiences disuse osteoporosis as a result of relatively stiffer implant materials carrying a portion of the JRF and thereby 'shielding' bone from normal, physiologic stress. To study our objective we employed a model of adaptive bone remodeling that affected change in bone density through stimulation of BMUs by disuse (i.e. insufficient loading). Because experimental evidence suggests fatigue microcracks in bone influence remodeling, damage as a stimulus of BMU activity was also included.

In comparing the potential for bone loss associated with stress shielding (Fig. 5), the interlocking stem with full bony ingrowth (bonded interface) was found to be a viable alternative to more common fixation techniques (namely, cement and press-fit with full bony ingrowth). Significant bone loss below the tibial tray (region 1) has been observed by X-ray absorptiometry [24]. In this region, our model predicted similar bone loss for both the cemented and interlocking stems. The former fixation technique caused the least amount of bone loss in the model, but cement suffers several drawbacks, such as potential thermal necrosis during polymerization, cement fatigue failure, and difficulty in removal of cemented components at revision. Our model also predicted that an interlocking stem without bony ingrowth (i.e. fluted or matted stem) did not result in less proximal bone loss when frictionless contact was assumed. A possible explanation is that in providing distal fixation with screws the tray of the interlocking stem is prevented from optimally loading the metaphysis.

To the best of our knowledge, the present study is the first to investigate remodeling around long stems with interlocking screw fixation. Nonetheless, the present model is consistent with the finding that trabecular bone loss in the metaphysis is greater than cortical bone loss in the diaphysis after TKA [24]. Furthermore, the present results are similar to those observed in long stemmed THA: stress shielding effects are greater for press-fit, cementless stems than cement stems with a smaller diameter [17].

Bone mineral density (BMD) depends on the rate at which mineral accumulates in newly formed bone. Much of this mineralization occurs soon after the osteoid is laid down; however, secondary mineralization may occur over years until this bone is remodeled [16]. Thus, BMD depends on the rate of bone turnover. Following TKA, elevated remodeling
caused by disuse decreases the mineral content of newly formed bone; therefore, bone loss as quantified by BAF cannot be directly compared to bone loss as quantified by BMD. Nonetheless, the predicted trend in bone loss following TKA can be compared to that of BMD reported by clinical studies. The present adaptive bone remodeling study found substantial loss in the first weeks after long stemmed TKA and gradual decrease in bone loss from 6 months to 5 years (Fig. 6(A)). For a group of 19 patients who had undergone cemented TKA, Seitz et al. [33] reported that a loss in tibial bone density (measured by computed tomography) was stabilized after 1 year. Also similar to the results of our study, they found that the cortices near the stem had thinned. For patients with a porous-coated anatomic prosthesis, Petersen et al. [29] reported a 7–20% decrease in bone density (measured with dual photon absorptiometry) below the tibial tray during the first 6 months but only an average loss of 22% at 3 years. The present study is consistent with these clinical studies in that most bone loss occurs early in the life of TKA.

In developing the simulation of tibial adaptation after long stemmed TKA, a change in daily loading activity levels and frequency as well as a time period for the occurrence of bony ingrowth were not modeled. One reason for performing TKA surgery is to restore knee function, so after recovery the activity level of the patient presumably increases. Yet, the present model did not include an increase in loading after TKA, and it did not include a decrease in loading to simulate patient inactivity prior to surgery because these are relative effects. In the case of interlocking fixation, the model could have under-predicted bone loss because a period of sole distal fixation while bony ingrowth occurs was not included. In addition, loss of bone for the non-bonded interlocking stem could have been over-predicted because frictionless contact was assumed.

There are also limitations with both the FE model and the theory of bone remodeling that were employed to achieve the objective of the present study. Plane strain assumptions, exclusion of out-of-plane muscle, ligament, and joint forces, simplified tibial geometry, and idealization of material behavior limit the characterization of the mechanical environment by the 2D FE tibial model. The symmetric loading protocol resulted in a symmetric diaphysis (Fig. 4(A)). Thus, the results reported here do not capture the effect of asymmetric loading. Yet, when the center of pressure of the JRF was shifted medially, the model predicted a medial metaphysis with denser trabecular bone and a slightly thicker cortex than the lateral metaphysis. While developed from the documented biological processes of bone remodeling, the model of adaptive bone remodeling operates with specific assumptions about relationships that have not been experimentally verified (e.g., the relation between Ac.f and damage or disuse). Nonetheless, the coupled models produced the principal morphologic features of the intact tibia and simulated proximal bone loss after TKA (Fig. 4) when given approximate physiologic loading conditions.

Advantages of the present model when compared to strictly strain-adaptive remodeling theories include: damage influences remodeling behavior, temporary increases in BAF as BMUs remodel affect bone stiffness, and changes in bone density depend on the surface area available for initiation of BMUs. However, our model only simulates bone remodeling, the coupled action of osteoclasts and osteoblasts, and does not provide a geometric modeling response to overloaded bone. Also, densification at the stem tip has been reported for stemmed joint arthroplasties, and bone hypertrophy is recognized as a source of a stress concentration that facilitates periprosthetic fracture [32]. Load transfer
from implant to bone by the distal screws may dissipate the high stresses at the stem tip that causes this hypertrophy.

The use of interlocking titanium screws could be a viable alternative to more common fixation techniques of long stems. They provide a means of fixing a small diameter stem, which helps to minimize stress shielding, and avoids the multiple problems associated with fully cementing a prosthesis. Using a damage- and disuse-based theory of bone remodeling, our model predicted that interlocking screws caused less bone loss than pressfit fixation and only slightly more bone loss than cement fixation. Also, results of this study suggest porous coating the interlocking stem for bony ingrowth may be advantageous if it prevents stress shielding associated with distal fixation.

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References


Fig. 1. A radiograph taken 7 years postoperative shows a fixed revision TKA with interlocking titanium screws that pass through prefabricated holes in a long stem.

Fig. 2. Implant components and an outline of a tibia (A) guided the development of the FE model. Regions (1–4) where values of model state variables were averaged are indicated on the mesh (B).
Stiffness, $E$ \( \rightarrow \) Loads, Geometry

\[ E = A \ast BAF^B \]

(i) BMU Activation, Frequency, $A_c.f$

\[ \frac{dD}{dt} = k_D \ast \Phi \]

(ii) $\Phi < \Phi_0$

\[ \frac{dDR}{dt} = D \ast A_c.f \ast R_s \ast Ar \ast F_s \]

(iii) $\Phi > \Phi_0$

Disuse, $\Phi < \Phi_0$

Strain, $S$

Damage, $D$

Damage Formation Rate, $dD/dt$

Damage Removal Rate, $dDR/dt$

Formation Rate, Removal Rate, $dD/dt$, $dDR/dt$

$E = A \ast BAF^B$

$\frac{dD}{dt} = k_D \ast \Phi$

$\frac{dDR}{dt} = D \ast A_c.f \ast R_s \ast Ar \ast F_s$

# of Filling and Resorbing BMUs, $N_F$ and $N_R$

Available Surface Area for Remodeling

BAF

Fig. 3. Stimulated by disuse and damage, BMU activation frequency dictates the adaptation of bone to its mechanical environment, which is characterized by principal strain [15].

Fig. 4. Comparing BAF distribution for the intact tibia at steady state (A) to that for the press-fit stem at 2000 days postoperative (B) reveals loss of bone between the tibial tray and stem tip.
Fig. 5. At 2000 days, bone loss as percent decrease in BAF of intact tibia occurs for each region and fixation type. The interlocking stem was either intimately bonded to bone or in sliding, frictionless contact with bone during simulation.

Fig. 6. Acquired from the simulation of the bonded interlocking stem as an example, two state variables plotted against simulated time reveal that after TKA (>0 days) BAF decreases rapidly within the first 6 months following TKA (A) because a state of disuse exists (B).