The Predictive Value of Stress Shielding for Quantification of Adaptive Bone Resorption Around Hip Replacements

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The presence of a femoral hip stem changes local mechanical signals inside the surrounding bone. In this study we examined the hypothesis that the eventual loss of bone can be estimated from the initial patterns of elastic energy deviation, as determined in FE models of the intact bone and the operated femur. For that purpose two hypothetical relations between elastic energy reduction and resorption were investigated. Their estimates of bone loss were compared to the results of iterative computer simulations. Two kinds of FE model were used, and in each stem stiffness and remodeling threshold (a measure of "biological reactivity") were varied. Provided that reasonable values of the remodeling threshold are assumed and that the stem is firmly bonded to the bone, we found that the difference between direct estimates and simulation models was 4 percent of bone loss. It is therefore concluded that initial patterns of elastic energy deviation give a reasonable indication of expected bone loss.

Introduction

The presence of a femoral hip stem reduces the mechanical signal level inside the surrounding bone (stress shielding), which can cause severe bone resorption (e.g., Bobyn et al., 1992; Huiskes et al., 1992). The process of bone resorption can be simulated in computer models by combining a mathematical description of adaptive bone remodeling with Finite Element (FE) models (Orr et al., 1990; Huiskes et al., 1992). These methods can play an important role in the design process of a prosthesis, especially since automatic FE-mesh generation from CAD drawings is becoming more common. For routine use in extensive parametric studies, such as encountered in design optimization, the present methods may use too much computer time. In addition, they do not allow the mathematical calculation of design sensitivities, which makes them less useful for efficient optimization routines and may slow down the optimization process.

In this study, we addressed the question whether knowledge of only the pre-operative and directly post-operative mechanical signal distribution would be sufficient to provide a reasonable assessment of the amount of bone loss. In other words, can the long-term amount of bone loss be predicted from the initial stress shielding effect of an implant? If so, then one FE iteration could replace a remodeling analysis typically requiring 20 iterations, saving 95 percent of CPU time.

Methods

We assumed that all bone will be resorbed when the mechanical signal S after insertion of the prosthesis is zero. Conversely, all bone will be preserved when the signal is equal to that prior to insertion of the prosthesis, the "natural" or reference signal S_{ref} . Around the reference signal a threshold level s was assumed (Carter, 1982, 1984; Frost, 1983, 1987; Huiskes et al., 1987, 1992; Nauenberg et al., 1993). Mechanical signal reductions that do not reach the threshold level will not cause bone resorption. Using an "influence func-

tion" $g(S)$, which has a value of one when the signal is zero and a value of (almost) zero when the signal is equal to the reference signal minus the threshold, the total resorbed bone mass M_r was obtained by integrating the product of original bone density ρ and influence function g over the total bone volume Ω . The resorbed bone mass fraction m_r was obtained by normalizing M_r to the original bone mass M :

$$m_r = \frac{1}{M} \int_{\Omega} g(S(\mathbf{x})) \rho(\mathbf{x}) d\mathbf{x} \quad (1)$$

where:

- m_r = resorbed bone mass fraction
- M = original bone mass
- Ω = original bone volume
- $\mathbf{x}$ = volume coordinate
- $g(S)$ = influence function giving resorbed bone mass fraction
- $S(\mathbf{x})$ = local value of mechanical signal
- $\rho(\mathbf{x})$ = local apparent bone density

Two influence functions were tested. The first function is the normal or Gaussian cumulative distribution function (Fig. 1, curve A):

$$g_1(S) = 1 - \frac{1}{\sigma\sqrt{2\pi}} \int_{-\infty}^s \exp\left[-\frac{1}{2}\left(\frac{\xi - \mu}{\sigma}\right)^2\right] d\xi \quad (2)$$

where:

$$\begin{aligned} \mu \text{ (mean)} &= 0.5(1.2 - s)S_{ref} \\ \sigma \text{ (standard deviation)} &= 0.1\mu \end{aligned}$$

The second influence function is a step function (Fig. 1, curve B)

$$g_2(S) = \begin{cases} 1 & \text{for } \frac{S - S_{ref}}{S_{ref}} < -s \\ 0 & \text{for } \frac{S - S_{ref}}{S_{ref}} \geq -s \end{cases} \quad (3)$$

Following Carter et al. (1989), Huiskes et al. (1989, 1992),

Contributed by the Bioengineering Division for publication in the JOURNAL OF BIOMECHANICAL ENGINEERING. Manuscript received by the Bioengineering Division August 10, 1993; revised version received September 16, 1996. Associate Technical Editor: S. A. Goldstein.

Fig. 1 Two choices for resorbed bone mass fraction g as a function of stimulus S and reference stimulus S_{ref} , in the presence of a threshold zone $s > 0$. Curve A is a Gaussian distribution function, and curve B is a step function.

and Weinans et al. (1993) we used the elastic energy per unit of bone mass, averaged over a daily cycle of n loading cases, as mechanical signal

$$S = \frac{1}{n} \sum_{i=1}^n \frac{U_i}{\rho} \quad (4)$$

where:

U = strain energy density ($\frac{1}{2}\sigma\epsilon$) for loading case i

ρ = apparent bone density

n = number of discrete loading cases

The validity of Eq. (1) was verified against computer simulations of the adaptive remodeling process (Huiskes et al., 1987, 1989, 1992). The validity of these simulations for assessing long-term bone density patterns was demonstrated relative to animal-experimental information (Weinans et al., 1993; Rietbergen et al., 1993) and human post-mortem information (Engh et al., 1992; Huiskes, 1993). We assumed that these simulations provide a good benchmark. The simulation method assumed a nonuniform reference value S_{ref} whose distribution was found by performing an FE analysis of the intact (nonoperated) femur (site-specific remodeling rule).

Two FE models were used. The first was two dimensional and represented a "straight stem" loaded with a bending moment of 1000 Nmm (Huiskes et al., 1987, Fig. 2(a)). The second model was three dimensional. The geometry and density distribution from the bone were determined from CT scans (Huiskes et al., 1992; Fig. 2(b)), and a cycle of three hip joint and muscle loading cases was used representing a daily loading pattern (Carter et al., 1989). For both models, a fully bonded "iso-elastic" ($E = 20,000$ MPa) and titanium ($E = 115,000$ MPa) prosthesis was modeled. For the two-dimensional model, the value of the threshold zone s was varied between 0.0 and 1.0. For the three-dimensional model, two values of the threshold zone s were used ($s = 0.35$ and $s = 0.75$) and bone loss was calculated in four zones. Zone 1 was bone around the proximal part of the prosthesis and zone 4 bone around the tip.

Results

Two-Dimensional Model. The normalized amount of stress shielding $(S - S_{ref})/S_{ref}$ had an almost constant value of -0.3 around the flexible stem and -0.9 around the stiff stem. In other words, the flexible stem reduced the mechanical signal for bone remodeling by 30 percent, and the stiff stem by 90 percent. In the remodeling simulation, varying threshold level s caused a sharp transition between nearly complete and almost no bone resorption (Fig. 3). For the iso-elastic prosthesis, this occurred at $s \approx 0.4$ and for the titanium prosthesis at $s \approx 0.9$. Using the step function (Eq. (3)), the estimated value of the

Fig. 2 Finite element models used: (a) Two-dimensional finite element model of a simplified bone/prosthesis configuration. (b) Three-dimensional finite element model of a bone/prosthesis configuration. The bone geometry and the bone density distribution have been based on CT scans. The three loading cases represent the daily loading cycle and have been derived from Carter et al. (1989).

bone loss fraction closely followed the value from the remodeling simulation over the complete range of the threshold level s . The Gaussian function (Eq. (2)) produced a more gradual curve, and underestimated bone loss.

Three-Dimensional Model. Contrary to the two-dimensional model, the normalized amount of stress shielding $(S - S_{ref})/S_{ref}$ was far from uniform (Fig. 4). Three remodeling simulations were performed, two using a titanium stem ($s = 0.35$ and $s = 0.75$) and one using an iso-elastic stem ($s = 0.75$; Huiskes et al., 1992). The Gaussian function gave close agreement between estimation and simulation in all regions, regardless of threshold level or prosthetic stiffness. The maximum deviation was 2 percent for total bone loss (iso-elastic stem), 8 percent at level 1 (iso-elastic stem), 12 percent at level 2 (titanium stem, $s = 0.75$), 4 percent at level 3 (titanium stem, $s = 0.35$ and $s = 0.75$), and 5 percent at level 4 (titanium stem, $s = 0.35$). With the step function, estimation and simulation only came close for $s = 0.75$. The maximum deviations occurred for the titanium stem and were 4 percent for total bone loss, 6

Fig. 3 The resorbed bone mass fraction obtained from remodeling simulations and estimated from the amount of stress shielding, as a function of threshold zone s and for two different stem moduli. The two-dimensional bone/prosthesis configuration was used.

Fig. 4 The normalized amount of stress shielding $(S - S_{ref})/S_{ref}$ in the midfrontal plane around an iso-elastic stem (left) and a titanium stem (right) for the three-dimensional configuration

percent at level 1, 3 percent at level 2, 6 percent at level 3, and 4 percent at level 4.

Visually, the bone density distributions calculated using remodeling simulation and step function around the titanium stem ($s = 0.75$) compared very well (Fig. 5). There were local differences, such as the resorption of proximal medial cancellous bone and the formation of a lateral bridge of cortical bone during the remodeling process, which were not predicted by the estimation procedure.

Discussion

Bone remodeling is a nonlinear process, during which altered mechanical signals change bone density patterns, leading in their turn to an altered local mechanical signal. Phenomena may therefore occur later that cannot be deduced from the initial situation, in particular around unbonded prostheses. But even when the prosthesis is bonded, resorption of bone in one location imposes higher loads on bone elsewhere, which may subsequently reduce the amount of bone lost. The step function (Eq. (3)) completely neglects this kind of nonlinearity. The Gaussian distribution curve (Eq. (2)) partially accounts for it, but bone apposition is neglected.

Nonlinearity depends mainly on the nonuniformity of the initial signal distribution, in association with the threshold level s . The simple two-dimensional configuration hardly showed any nonlinearity, and the estimate of bone loss using the step function came closest to the remodeling simulation. In this case the Gaussian curve did not perform well at all. Conversely, for the three-dimensional configuration the Gaussian curve gave a close approximation to the remodeling simulation in all cases. The step curve gave only close approximations of bone loss for the larger threshold level of 75 percent. Apparently, such a large threshold reduces nonlinearities to a sufficiently low level.

The notion of a threshold was first introduced by Carter (1982, 1984) and Frost (1983, 1987). Clinical support came from observations by Maloney et al. (1989), Engh and Bobyn (1988), and Nauenberg et al. (1993). Theoretical support came from Huiskes et al. (1987, 1992) and Weinans et al. (1993), who showed that a threshold zone is required for a realistic

Fig. 5 The density distribution in the midfrontal plane around a titanium prosthesis, assuming a threshold zone $s = 0.75$. Left: obtained from a remodeling simulation. Right: predicted using the step function (Eq. (3)).

simulation of the remodeling process. The value of the threshold level is not known. Engh and Bobyn (1988) estimated that a loss of 30–45 percent of the bone stresses (or 75 percent of the strain energy density) seems tolerable, which would suggest a level of 0.7–0.8. Huiskes et al. (1992) found that a threshold zone of 0.75 was required to predict realistic resorption patterns. It thus seems that humans have a relatively large threshold level, which means the condition is met for using the step function to predict bone loss.

In conclusion, if the prosthesis is bonded and a relatively large threshold describes the reactivity of bone, an estimate of percentages of long-term bone loss similar to that obtained from more extensive simulation analyses can be obtained from the initial extent of stress shielding by using a simple step function to relate stress shielding to bone loss.

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