JOINT LOAD ESTIMATION BASED ON BONE DENSITY AND A CONTACT MODEL APPLIED TO THE PROXIMAL FEMUR OF A CHIMP

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INTRODUCTION

Mechanical loading significantly influences the internal density distribution and external morphology of bone. The tissue is continually remodeled toward a density distribution that maintains a certain level of localized mechanical stimulus. Based on the assumption that a given bone has been subjected to mechanical loading conditions that maintain a consistent level of local mechanical stimulus, the bone density distribution should provide direct information on that consistent loading applied to the bone [1, 2]. Frost [2] relates the amount of cross-sectional bone tissue just below a joint surface to the total joint load. Fischer et al [1] used a linear computational method to estimate the dominant loading patterns on the proximal femur of a human. Finite element models were made of the human proximal femur, and a variety of parabolic pressure distributions were applied as nodal forces. An optimization routine was then used to select and scale the applied quadratic pressure distributions, based on the remodeling theory of Beaupre et al [3], so that they created the desired level of local mechanical stimulus throughout the bone.

The objective of this study was to extend the linear computational method to allow for joint contact. By modeling the contact between the acetabulum and the femoral head, we apply load distributions that more closely resemble in-vivo loading. This should improve the estimate of joint loading patterns based on the bone density distribution.

THEORETICAL FRAMEWORK

The current theory assumes that the local bone tissue remodels to maintain a local attractor state stress stimulus, $\psi_{b_{AS}}$. The applied tissue stimulus, ψ_b, is based on the local strain energy density of the bone tissue [3] and is defined as

$$\psi_b = \left[\sum_i n_i \bar{\sigma}_i^m\right]^{1/m} \left(\rho_c/\rho\right)^2 \qquad (1)$$

where n_i is the number of cycles of load type *i* per day, $\bar{\sigma}_i = \sqrt{2EU}$ is the continuum level effective stress for load case *i*, U is the strain energy density for the load case *i*, E is the modulus of elasticity of the local bone tissue, *m* is a stress exponent that weights the relative importance of load magnitude, and the squared ratio of cortical density over local apparent density scales the continuum values to the tissue level.

Figure 1. CT Scan of the Chimp Femur and Acetabulum (Coronal Slice)

If a bone is assumed to be in remodeling equilibrium, the applied loads have created a local stress state equal to the local attractor state stress stimulus at every point. The right hand side of Equation (1) is equal to $\psi_{b_{AS}}$ except where the bone is fully cortical and cannot be further remodeled. If the material is linear elastic, then we can apply a set of scalable loads to the model. We can then construct a global optimization function similar to that of Fischer et al [1], to be minimized with respect to the magnitudes of the scalable loads,

$$\Phi(x_i) = \frac{1}{2\ \text{npts}} \sum_{i=1}^{\text{npts}} \left\{ \sum_{j=1}^{\text{nloads}} 1 - \frac{\left[\sum_{\text{day}} n_i \bar{\sigma}_j^m(x_i)\right]^{1/m} \left(\frac{\rho_c}{\rho}\right)^2}{\psi_{b_{AS}}} \right\}^2 \qquad (2)$$

where *nloads* is the number of candidate load cases, *npts* is the number of integration points in the finite element model, x_i is the magnitude of the current load case, and $\overline{\sigma}_j(x_i)$ is the effective stress at node *j* for the current magnitude of load case *i*.

The load magnitudes that minimize Equation (2) represent an estimate of the daily loading conditions. Due to the nature of the method and the density data, these loads indicate dominant loading patterns and not absolute or dynamic in vivo loads.

METHODS

We applied the load estimation method to the proximal femoral head and acetabulum of a chimpanzee (Fig. 1). A two dimensional coronal slice of the femur and acetabulum was reconstructed from three-dimensional single energy CT data. The grayscale pixel values were then converted to density values ranging from fully cortical bone (1.92 g/cc) to a minimum value of (0.05 g/cc) which corresponds to minimal bone. The external geometry of the femur was then used to create a finite element mesh and densities were assigned to the element nodes and interpolated over the elements. We predicted joint loading patterns based on these models using both linear load estimation and the joint contact method.

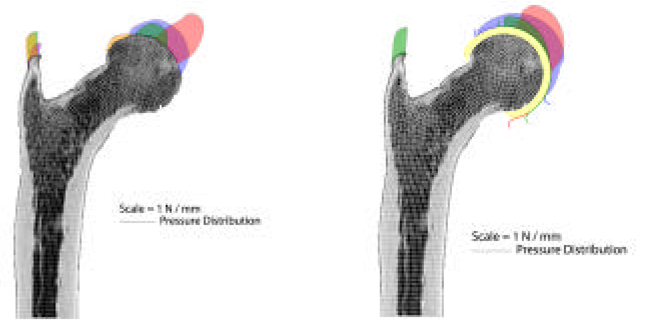
Linear Computational Model

Quadratic pressure distributions spanning 25, 20, 15 and 10 nodes were applied at regular increments over the surface of the femur. Loads were also applied to the trochanter and were paired with pressure distributions on the femoral head. In all, 51 separate load cases were considered. A finite element analysis was then performed using ABAQUS 6.3-1© (Hibbitt, Karlsson & Sorensen, Inc., Pawtucket, R.I.) and the objective function in Equation (2), was minimized with respect to the magnitudes of the applied quadratic pressures using a non-linear least squares optimization routine from NAG© (Numerical Algorithms Group, Downers Grove, Il). The results of the linear method are shown in Figure 2A, where the shaded areas represent the predicted distributions.

Contact Model

A finite element mesh of the femur was constructed in the same way as for the linear model. In addition, the acetabulum was digitized and modeled in ABAQUS© as a rigid analytical surface. Since the chimpanzee femur and acetabulum were obtained from the collection of skeletal animal remains from the University of Kansas Natural History Museum, the thickness of the cartilage was not known. To approximate this thickness, circles were fit to the femoral head and the surface of the acetabulum. The difference in the radii was then used as the total thickness of the articular cartilage from both surfaces.

The acetabulum was positioned so that the center of the acetabular circle coincided with the center of the femoral head circle. The acetabular surface was rotated at 10º increments about the femoral head and loads applied at each increment both perpendicular to the surface and at angles 45º clockwise and counter-clockwise from the perpendicular load. As in the linear model, the head loads were matched to trochanter loads. Since changes in external loads create changes in the contact profile between the cartilage and the acetabulum, we cannot simply scale the loads. Instead, a quasi-linear method was developed in which the linear optimization was used to determine the direction in which the load scaling factors should be adjusted. The load scales were not allowed to change by more than 10% after the optimization. The input files were then updated and the analysis performed again. The analysis was stopped when the optimization routine selected scaling factors of either all 1.0 or zero for all load scales, with a tolerance of 0.01. The results from the contact method are shown in Figure 2B.



A B

Figure 2. Finite element models with the predicted loading patterns for both linear (A) and joint contact (B) methods

RESULTS

The density-based load estimation method was successfully implemented in a contact model. Due to the close fit between the acetabulum and the cartilage surface on the head of the femur, the change in the contact profile as loads were scaled was minimal. As a result the non-linearities were not pronounced, and the quasi-linear method converged relatively quickly. In a more incongruent joint, the non-linearities would be more noticeable.

Similar trends in the load distributions can be seen between the linear method and the contact method (Fig. 2). Both select one large load distribution and two smaller load distributions at about the same locations on the femoral head. In addition, the linear method selects a load distribution on the superior-lateral region of the femoral head, and two additional trochanter loads.

DISCUSSION

All the load distributions in the linear method are quadratic. The contact method produces pressure distributions that are varied, with the large distribution similar to a quadratic distribution and the two smaller ones considerably different. This is due to the method selecting loads that are not always applied perpendicular to the contact surfaces, a situation that would be expected in-vivo.

While more complex than the linear method, the contact method has important advantages. The contact method allows us to associate dominant loads with a given direction of joint loading and the relative position of the femur and pelvis. The contact method will also allow a more straightforward application of potential load cases in 3-D density-based load estimation.

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