Development of Strain Adaptive Finite Element Model for the Ulna

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Introduction

The use of orthopaedic implants in the treatment of damaged or diseased joints results in a phenomenon known as "stress shielding", in which loads regularly borne by mineralised bone tissue are instead diverted through the stiffer prosthetic. The response of the bone tissue is to resorb in the area surrounding the implant, which may be detrimental to long term implant survivability [1]. Material models capable of reproducing this behavior can thus provide valuable input in the field of orthopaedic implant design.

One approach to describing this behavior assumes that bone density is regulated by strain energy, described by Equation 1 [1]:

$$\frac{d\rho}{dt} = \begin{cases} \left({U}/{\rho} - (1+s)K\right) & (1+s)K < {U}/{\rho} \\ 0 & (1-s)K < {U}/{\rho} < (1+s)K \\ \left({U}/{\rho} - (1-s)K\right) & {U}/{\rho} < (1-s)K \end{cases} \qquad 1)$$

In this model, the body strives to maintain a homeostatic strain stimulus value K (strain energy per unit mass, J/g) within bone tissue. Any externally induced stimulus (strain energy density U (J/cm^3) per mass density ρ, (g/cm^3)) greater or less than this value by a given threshold $s\times K$, ($0<s<1$) triggers a respective positive or negative rate of bone formation ($d\rho/dt$, g/cm^3/time), stiffening bone in areas of high strain, and vice versa. One approach to validating this technique is to initialize a finite element (FE) model of a bone with a homogeneous density distribution, subject it to an assumed *in-vivo* loading environment, and test the hypothesis that the density distribution will converge to that of a physiological bone.

This validation method has been implemented in FE analyses of the proximal femur [2] and glenoid [3], with values for *K* and *s* ranging from 2.5E-3 – 4.0E-3 and 0.35 – 0.75, respectively. This indicates that optimal parameter assignment may vary with anatomical location and knowledge of the loading environment. This procedure has yet to be applied to the distal ulna, and as such, the purpose of the present study is to determine optimal values for the parameters *K* and *s* in the ulna.

Materials and Methods

A 3D computer model of a right ulna (male, 76 years) was created using Mimics® (Materialise, Leuven, Belgium) and μCT data, and then meshed with 2nd order tetrahedral elements (characteristic length of 0.7 mm) using Abaqus® (Providence, RI). The strain-adaptive material model was executed using custom-written code implemented via the user subroutine UMAT. A program developed in-house was used to derive density values for each mesh element from the μCT scan, creating a reference model with a physiological density distribution. Meshes subjected to the remodeling procedure were set to a homogeneous density of 0.8g/cm^3. Minimum and maximum density values of 0.01 g/cm^3 and 1.85 g/cm^3 were imposed for all models, representing hollow space and cortical bone, respectively.

A relation between elastic modulus and bone density, previously validated for the ulna, was used to assign linear elastic material properties to each mesh element [4]. A 20N medially-directed load was applied to the distal ulnar head, with the proximal end of the ulna fixed. This load was applied over 100 iterations, with Equation 1 acting on all mesh elements following each load application.

Three constant values for both *s* (0.35, 0.55, 0.75) and *K* (2.5E-5, 2.5E-4, 2.5E-3J/g) were used. A site-specific (SS) approach for *K* was also tested, where each mesh element was assigned a unique *K* value calculated using the strain field induced in a model of the ulna with physiological elastic properties.

All data analysis was limited to the diaphyseal region of the ulna within the applied bending plane, as the strain distribution due to a single bending load was not representative of the physiological case. Model accuracy was evaluated using the root-mean-square-error (RMSE) between the density values calculated by the FE simulation and those derived from the μCT scan. Percent error was calculated to evaluate any biases, and density plots were created to visually compare physiological and computed density distributions.

Results

Table 1 lists the RMS and average percent error between calculated and scanned mesh element density values in the region of interest after 100 iterations. The models using s = 0.35 and 0.55 with K = 2.5E-4 J/g had the smallest RMS error, and were run for a further 100 iterations, resulting in RMS and percent error values of 0.46 g/cm^3 and -15% for s = 0.35, and 0.46 g/cm^3 and -10% for s = 0.55 (both with K = 2.5E-4 J/g). Figure 1 plots the density distribution for the model using s = 0.35 and K = 2.5E-4 J/g after 200 iterations and the physiological case.

Discussion and Conclusion

The parameter combination of s = 0.55 and K = 2.5 E-4 J/g resulted in the smallest RMSE value (0.46 g/cm^3) and absolute value of average error (-10 %) of all other parameter combinations, displayed steady behavior after 200 iterations, and was thus judged to produce the most physiologically accurate density distribution in the ulna. These results compare well with those calculated for the glenoid, which reported an RMSE of 0.49 g/cm^3 [3] using a site-specific value for K. In Figure 1, important bone structures in the diaphysis of the ulna are apparent, including a dense cortical shell and central medullary canal.

The assumption of a continuous, isotropic model for bone tissue ignores the trabecular structure in the epiphysis of the ulna, which have been shown to contribute to orthotropic behavior of bone [5]. Remodeling procedures have been developed that re-orient individual trabeculae with principal stress trajectories; however, these models are much more computationally expensive than the present model, and impractical as a design tool for orthopaedic implantation [6]. Furthermore, in the case of arthroplasty of the distal ulna, the epiphysis is removed, leaving the largely cortical diaphysis, for which the isotropic assumption has been shown to be adequate [4].

The 20 N bending load was chosen based on unresisted forearm rotation of the ulna [7], and is therefore on the low end of what might be experienced *in-vivo*. Thus, the parameter values selected as optimal for this study do not necessarily have any connection with human physiology; however, so long as the parameters are used consistently, the model may still be used in the evaluation of orthopaedic implant designs.

In the present study, a strain-adaptive material model for bone was optimized for the ulna. This model performed comparably to a similar analyses of the glenoid, and was able to reproduce the cortical shell and medullary canal observed in the ulna and other long bones. This technique shows promise as a tool to evaluate the temporal effects of orthopaedic implants design in the distal ulna.

References

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Table 1 – RMS and average % error in density as calculated after 100 iterations of the strain-adaptive model and acquired from the μCT scan.

K (J/g)	*2.5 E-3*			*2.5 E-4*			*2.5 E-5*			*Site-Specific K*		
s	*0.35*	*0.55*	*0.75*	*0.35*	*0.55*	*0.75*	*0.35*	*0.55*	*0.75*	*0.35*	*0.55*	*0.75*
RMSE (g/cm^3)	0.88	0.8	0.74	0.44	0.45	0.51	0.61	0.60	0.59	0.76	0.79	0.82
% Error	40	39	39	-12	-6	4	-61	-57	-49	32	29	27

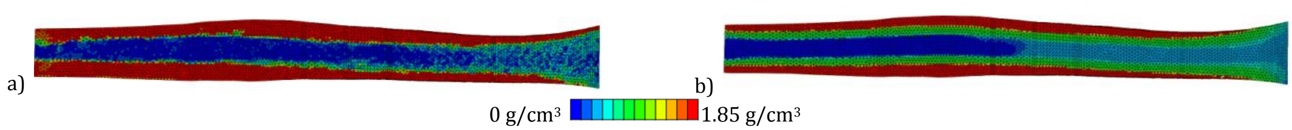

Figure 1 – The density distribution of the ulna in the a) physiological condition and b) as computed using *K*=2.5E-4 and *s*=0.55 after 200 iterations.