Endosseous implants have been used successfully for a variety of applications in dentistry\textsuperscript{1,2} and orthopedics.\textsuperscript{3} However, the scientific basis for the rigid ("osseointegrated") interface has developed slowly\textsuperscript{4} because of an incomplete understanding of relevant bone physiology.\textsuperscript{5,6} The processes involved in bone adaptation to an endosseous implant have been described.\textsuperscript{7} The healing response includes the formation of calluses and a mechanism for generating a vital interface between implant and bone. Rapid remodeling of bone near the interface has been shown consistently.\textsuperscript{5} It was expected that this rapid turnover would lead to a change in bone mechanical properties. Recently, microhardness of different bone types in various regions surrounding an implant have been evaluated.\textsuperscript{8} These data suggest that bone near the implant has distinctly different mechanical properties than cortical bone away from the implant. Also, it has been reported that endosseous implants bond to bone. The exact nature of this bond is unknown, but it is likely a combination of mechanical interlocking and intermolecular forces. These adaptational responses and bonding conditions at the implant-bone interface are certain to affect mechanical parameters\textsuperscript{9-11} (e.g., maximum and minimum principal strains and strain energy density) that are typically used to analyze loaded biologic structures. However, limited information is available on how these mechanical parameters change during healing and long-term adaptation. The purpose of this study was to use finite element (FE) methods to isolate the effects of callus formation and bonding on the mechanical environment in implant-supporting bone.
Materials and Methods

Three convergent sets of FE models were developed, each simulating a stage of healing and adaptation (Figs 1 and 2). The model geometry was developed for a study in progress in which dental implants were placed for a 12-week period in the midfemoral diaphysis of mature hounds. Figure 1 is a cut-away schematic showing model dimensions.

The first model, nonadapted (NA), is an immediate post-placement simulation (Fig 2a). The implant in this model is surrounded by cortical bone that has not had the opportunity to adapt. The second model, adapted (AD), represents bone that was allowed to heal for 12 weeks (Fig 2b), ie, one canine remodeling cycle.13 This is the waiting period after which provisional loading commences in clinical situations. The implant was surrounded by new bone adjacent to the implant and by calluses on the periosteal and endocortical surfaces. The three layers of new bone were meshed to represent the rapid remodeling gradient that is seen within about 1 mm of the interface. Each layer was given different material properties. The endosteal callus was meshed to fill the space between the endocortical bone surfaces and the implant. The third model, long-term adaptation (LA), is a hypothetical case simulating resorption of callus that accompanies long-term adaptation (Fig 2c). This model is identical to AD except that it lacks calluses. Change in cortical bone dimensions related to expected bone modeling during long-term adaptation was not simulated.

Bonded versions of NA and LA consisted of 5,344 elements and 6,340 nodes. The corresponding unbonded models had 6,106 elements and 7,012 nodes. Bonded AD had 7,104 elements and 7,938 nodes. Unbonded AD had 8,640 elements and 9,474 nodes. All models were generated through PATRAN (PDA Engineering, Costa Mesa, CA) and solved through ABAQUS (Hibbit, Karlson and Sorensen, Providence, RI). All elements were eight-noded bricks. In each model, two interface conditions were imposed. A model was either entirely unbonded or fully bonded. Absence of bond between implant and bone was simulated with contact pairs.14 These elements transmit compressive stresses but allow for separation under tension and sliding under shear. All unbonded models were idealized as frictionless. A cantilever bending load of 100 N at 13 mm from the periosteal side of the cortical bone was applied to each of the three models. This approximates the distance from the occlusal contact of a maxillary incisor to the crest of the alveolar bone.15 Fixed displacement boundary conditions were applied to the undersurface of the cut cortical bone. In all instances, the same boundary and load conditions were imposed.

Material properties (Table 1) of orthotropic canine cortical bone were estimated from the literature.16 Elastic moduli in the three orthogonal directions were based on ratios of $E_{rad}$:$E_{tan}$:$E_{long}$ = 0.65:0.8:1.16 No estimates for the elastic moduli of the calluses and implant-adjacent adapted bone are available. Thus, material property distribution was approximated from the microhardness gradient previously demonstrated.8 Poisson’s ratio of 0.34 was chosen for all bones. All bone types were considered to be linear, elastic, and homogenous. Material properties for titanium were $E$ = 104 GPa and $\nu$ = 0.36.17

It is important to emphasize that the geometry and mechanical properties used in these models simulate implants in dog femoral cortex. Because of the vast differences in the thickness, geometry, porosity, vascularity, and mechanical properties of the bone, no direct comparisons of results can be made to implants placed in the maxilla or mandible. Such differences exist between the anterior and posterior parts of each jaw, thus making within-jaw
comparisons difficult to generalize. Thus, in this study, only relative comparisons are made between models.

The peak values of four mechanical parameters—minimum principal strain (MIN), maximum principal strain (MAX), strain energy density (SED), and maximum shear strain (MSS)—were obtained from the FE models. In this study, MIN was always compression and MAX was always tension. The nodes along the length of the implant were divided into seven groups that extended from the periosteal surface to the endocortical surface of the bone for NA and LA. Because the calluses were present, there were 16 such groups for AD. Each group consisted of 17 nodes about the circumference of the implant’s interface. As an example, the desired values of strain energy density (SED) in bone were obtained as follows. In each group along the length of the implant, the peak value (SED$_{p}$) of strain energy density from the 17 circumferential nodes was identified. For comparisons, the highest value of strain energy density (SED$_{p}$max) among the 7 (NA and LA) or 16 (AD)
peak values thus obtained was used. For further comparisons, the highest peak values of the mechanical parameters in each of these models was normalized with the corresponding highest peak value of unbonded NA (e.g., SED_pmax in AD/SED_pmax in NA). NA was chosen because it corresponds to the time that implants are placed at initial surgery and forms the baseline from which comparisons can be made. In addition to comparisons of peak parameter values, their locations were also investigated.

Results

Healing, as defined herein, consists of two processes: (1) formation of calluses and new bone around the implant (modeling and remodeling events), and (2) formation of a bond between implant and bone. In laboratory experiments, the latter is indicated by an increase in interfascial bond strength. The effects of healing are depicted by comparing the unbonded nonadapted model (NA) with the adapted model (AD) in Figs 3 and 5. Calluses, new bone around the implant, and full bond between implant and bone decrease MIN_pmax by nearly one order of magnitude (8.3 fold). When unbonding at the interface of AD model was simulated, MIN_pmax decreased by only 3 times when compared to unbonded NA. Similarly, when unbonded NA is compared with unbonded or bonded AD, a dramatic decrease in MAX_p, SED_p, and MSS_p is also apparent (Figs 3b through 3d and 5).

Fig 3a  Peak values of minimum principal strain (MIN_p) at various node groups along the length of the implant normalized to the highest peak (MIN_pmax) for unbonded NA. B (solid symbols) and U (open symbols) indicate bonded and unbonded, respectively, in Figs 3a to 3d.

Fig 3b  As in Fig 3a, but for the maximum principal strain (MAX_p).
The importance of the calluses was assessed by comparing AD with LA. Callus resorption and lack of a bond increased $\text{MIN}_p_{\text{max}}$ by 6.6 fold (Figs 3a and 5). In contrast, $\text{MIN}_p_{\text{max}}$ in the bonded cases increased 2.7 fold. This is also illustrated when various other mechanical parameters are compared (Figs 3 and 5). Values of parameters for LA (both bonded and unbonded) are higher than their corresponding values in AD.

Development of an interface bond decreased $\text{MIN}_p_{\text{max}}$ 6.2 fold in NA and 6.4 fold in LA. However, bonding the implant to bone in AD decreased $\text{MIN}_p_{\text{max}}$ only 2.6 times. For all four mechanical parameters, bonded AD had the lowest absolute values (Figs 3 and 5). Bonding decreases the values of all mechanical parameters investigated as compared to the values for their respective unbonded models.

The effects of healing, bonding (interface conditions), and callus resorption on peak maximum shear strain ($\text{MSS}_p$) in bone followed the same trend as $\text{MIN}_p$. However, the peak values of strain energy density ($\text{SED}_p$) were greatly affected by some of these variables of healing. For example, when comparing unbonded NA with unbonded AD, there was a 20-fold decrease in $\text{SED}_p_{\text{max}}$ (Figs 3c and 5). Peak values of maximum principal strain ($\text{MAX}_p$) were less affected by healing for some of these comparisons. For example, $\text{MAX}_p_{\text{max}}$ in AD was 3.1 fold less than in unbonded NA.

**Fig 3c** As in Fig 3a, but for strain energy density ($\text{SED}_p$).

**Fig 3d** As in Fig 3a, but for maximum shear strain ($\text{MSS}_p$).
Figures 4a to 4c depict the location of $\text{MIN}_p$ for unbonded NA and bonded AD along the circumference of the implant for the various node groups. These simulated the two extreme conditions of adaptation. In unbonded NA, the highest value of $\text{MIN}_p$ was located at the endocortical surface close to the plane of loading (Fig 4b). However, some of the other $\text{MIN}_p$ values in groups of nodes were located at $\Theta$ other than 0 degrees and 180 degrees. In bonded AD, $\text{MIN}_p$ was located essentially at $\Theta$ equal to 0 degrees or 180 degrees, while the highest value was located in the periosteal callus away from the plane of loading (Fig 4c). However, the value of $\text{MIN}$ at $\Theta = 0$ degrees was not very different (thin bar in Fig 4c). Similar trends were seen for $\text{MAX}_p$, $\text{SED}_p$, and $\text{MSS}_p$.

Discussion

Finite element models in this study were used to provide relative estimates of the mechanical parameters under consideration. These estimates depend on the assumed material property values. Overestimation by studies using ultrasonic techniques \(^{18}\) was taken into account in the selection of the elastic and shear moduli of canine cortical bone. \(^{16}\) It is well known that bone is anisotropic and viscoelastic. Thus a limitation of this study is that bone was modeled as having isotropic and linear elastic properties, except for cortical bone, which was considered orthotropic. Others have also considered woven bone to be isotropic. \(^{19}\) However, it is expected that with maturation of the callus by lamellar compaction and mineralization, it will develop anisotropy. \(^{19}\) The only available estimate for the elastic modulus of fracture callus is through...
correlations with acoustic impedance.20,21 However, it is possible that the healing and final resorption processes of calluses around endosseous implants are different from those in fracture callus, thus negating extrapolations. A distinct gradient of increasing microhardness has been shown to exist within 600 µm of the implant interface.8 Also, a relationship between microhardness and elastic modulus has been demonstrated.22,23 Bone adjacent to the endosseous implants remodels rapidly, and its osteonal nature is distinct.7 Insufficient information was available to model the calluses and adjacent implant bone as anisotropic, although there are some observations of preferential orientation of osteons in this bone.6,7 Therefore, based on previous histologic findings2 and microhardness values,8 estimates for the Young's modulus of bone within 1 mm of the implant interface and calluses were made (Table 1).

Canine long bones are frequent sites for the experimental placement of endosseous implants.6,24 While in this study a load was applied at a specific location to simulate intraoral conditions, canine cortical bone material properties were used. This study has developed from another canine FE model that is currently being examined. While information on the orthotropic properties of the mandible is available, such data on the alveolar cortex are not. It is emphasized that in the present study only relative comparisons between models were made.

Considerable controversy exists in the literature on the ability of bone to bond to implant materials. Biocompatibility of the implant material, surface roughness, topology at the interface, and test methods are frequently cited as sources of this controversy.24-26 It is generally accepted that calcium phosphate and glass ceramics are bioactive27 and make direct contact with bone.28 On the other hand, titanium and its alloys are biocompatible materials but lack high interfacial bond strengths when compared with bioactive materials.24 No clear-cut relationship between surface roughness and bond strength exists. However, a general relationship has emerged that indicates that smooth surfaces promote the formation of thick fibrous encapsulation, and rough surfaces favor more intimate bone integration.29 It is possible that rough surfaces would limit micromotion at the interface; however, a number of other variables30 may be important. The titanium implant modeled in this study was nonthreaded. Based on a review of the literature,31,32 it is unlikely that the interface possesses a high shear bond strength. Thus, two extremes were modeled: fully bonded and unbonded. It is likely that in vivo bonding conditions lie between these two extremes.

Only cantilever bending load conditions were simulated. The results indicate that if bonding is assumed, MINpmax around the implant decreases by approximately 6.5 fold for NA and LA when compared to their respective unbonded models. The lack of a bond could result in relatively high strains, for example, in unbonded LA. Our results concur with those of Siegele and Soltesz33 and Hipp et al,34 but contradict the finding of Rieger et al.35 Siegele and Soltesz33 demonstrated a 2- to 5-fold increase in the values of maximum compressive stress adjacent to an unbonded implant in comparison with one with a full bond, using a 100 N axial load. Hipp et al34 empha-
size the importance of interfacial bonding assumptions. Their data suggest that an unbonded model would have larger values of maximum stress and greater motion than the bonded model. However, based on results of stress distribution, Rieger et al found no biomechanical advantage to having a fully bonded interface when compared to one that has bone closely adapted to the implant. Clift et al suggest that the results of Reiger et al may be explained by the fact that a nonsophisticated numeric algorithm was used to describe the allowable contact between implant and bone.

Intuitively, we expected $\text{MIN}_p$ to be located along the plane of loading ($\theta = 0$ or 180 degrees) of unbonded NA. However, toward the center of the cortical bone, $\text{MIN}_p$ values were located away from the plane of loading. In contrast, $\text{MIN}_p$ values for bonded AD were located along or near the plane of loading. This again emphasizes the differences between these two models and the effects of bonding. Analogies between fracture healing and healing at the implant interface have been made. The interfragments strain hypothesis attempts to provide a unifying theory to explain the relationship of strain, fracture immobilization, and tissue response. Gardner et al demonstrate that fracture motion triggers callus proliferation. This in turn results in increased stability. Carter et al proposed that the increased constraint may function to decrease strains and to allow bone formation to proceed. Results of the present study demonstrate the importance of calluses in decreasing various mechanical parameters both in unbonded and bonded situations. Calluses decrease $\text{MIN}_p \text{max}$ by approximately 3 to 7 fold, depending on the interface bonding condition. Brunski suggests that the stability of an implant is paramount in the development of a mineralized versus nonmineralized interface. Thus, reduction in strains by the callus may provide an environment for regeneration.

Endosseous implants are routinely placed with a two-stage procedure in accordance with Brånemark’s philosophy. However, histomorphometric data from screw-shaped implants suggest that much earlier or even immediate loading may be possible. It has been suggested that improved screw-shaped implant designs with greater retentiveness would decrease micromotion to such a degree that regeneration of bone may be possible even with early loading. However, in this study of a nonthreaded implant, the results clearly demonstrate large differences in strains of the NA and AD model. It would be interesting to develop a similar FE model of a threaded implant. Also, since the maxillary sites for implant placement have thinner cortical bone for supporting implants, the effect of cortical bone thickness on various mechanical parameters should be investigated. Currently, the effects of compressive and torsional loads on the three bonded models are being examined, and comparisons between various mechanical parameters will be made.

**Conclusion**

Healing response subsequent to implant placement is characterized by formation of calluses, rapid remodeling of bone adjacent to the implant, and an increase in interfacial bond strength. The dramatic effects of these responses on various mechanical parameters are demonstrated in this study. These parameters are considered to be determinants of bone response to mechanical loads. The results suggest the importance of the stabilizing roles provided by the callus and development of a bond during the critical phases of bone healing and long-term adaptation.

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