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THE PLATED FEMUR: RELATIONSHIPS BETWEEN THE CHANGES IN BONE STRESSES AND BONE LOSS*

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Calculations were made of the alterations in the *in vivo* cyclic bone stresses due to the application of various plates on the canine femoral shaft. The plate configurations analyzed were those used by previous investigators when studying the influence of plating on bone remodeling. The magnitude of the reduction in the loads borne by the bone tissue and the degree of shift in the bone stress neutral axis during the stance phase of gait was influenced by the geometry of the plate, the plate elastic modulus, and the location of plate application. From a correlation of the calculated alterations in bone stresses with the resulting measured changes in bone mass, it appears that bone remodeling is very sensitive to small changes in cyclic bone stresses. Changes in cyclic bone stresses of 1 MPa (less than 1 percent of the ultimate strength) can cause measurable differences in bone remodeling after a period of a few months.

Key words: bone loss; bone stresses; plate fixation

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The application of a plate to a long bone which has been fractured can establish excellent alignment and allow healing to proceed with a minimum of callus formation. After the fracture has healed, however, the plate transmits a substantial portion of the loads which are normally transmitted by the bone. Extensive bone remodeling and a net bone loss may occur which many researchers attribute to the changes in *in vivo* bone stresses (Uhthoff & Dubuc 1971, Tonino et al. 1976, Akeson et al. 1976, Moyen et al. 1978, Bradley et al. 1979, Pilliar et al. 1979). Additionally or alternatively, bone loss and remodeling may be influenced by the vascular insult introduced by the fracture and/or the presence of the plate (Jacobs et al. 1980). Experimental studies indicate that, in general, greater bone remodeling and bone loss is observed when stiff plates are used than when more flexible plates are used. However, no data have been presented evaluating the magnitude of the changes in stress and correlating these with the metabolic response.

Previous studies

Ten weeks after applying a 4-hole Richards compression plate to the lateral aspect of canine femora to stabilize a transverse osteotomy, Uhthoff & Dubuc (1971) noted three important changes which occurred in the bone: 1) osteopenia was pronounced in the cortex directly underlying the plate, 2) the shaft calibre was reduced, presumably due to periosteal resorption, and 3) new bone laid down in the fracture region did not exhibit the well organized structure of the old bone which previously occupied the area.

Tonino et al. (1976) studied bone remodeling after applying plates of different moduli on the lateral aspect of intact canine femora bilaterally.

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They compared 316L stainless steel AO plates to plastic polytrifluoromonochloroethylene (PTFCE) plates. The dogs were sacrificed 10 to 18 weeks after plate application. Histologically, great differences were seen between the bones plated with AO steel plates and those with PTFCE plates. In the AO plated femurs there were areas of extensive intracortical porosity. There was evidence of new bone formation in many of the large cavities, but edge sclerosis indicated that much of this new bone formation had already ceased in a number of the cavities. In the PTFCE plated femora, the resorption cavities were much smaller and bone loss was not as extensive.

In contrast to the findings of Uhthoff & Dubuc, Tonino et al. observed that resorption occurred mainly on the anterior and posterior aspects of the bone rather than directly under the plate in the lateral cortex. Three point bending tests of the intact harvested whole bones showed that the steel plated femora were significantly weaker than the PTFCE plated femora. Furthermore, the elastic modulus of small bone specimens cut from the steel plated femora was less than that of specimens taken from the PTFCE plated femora, suggesting that more extensive remodeling and/or a greater increase in porosity may have adversely affected the bone material properties.

Akeson et al. (1976) and Woo et al. (1976) also investigated bone remodeling following the application of fracture plates with different stiffnesses on the anterior aspect of the intact canine femur, comparing Vitallium plates to graphite fiber methyl methacrylate (GFMM) plates. The plated bones were subjected to mechanical, quantitative histological, and cortical thickness studies 9 and 12 months after plate application. Significantly greater bone loss was observed in the femora plated with the Vitallium plate than the GFMM plate. This bone loss was primarily manifested by a reduction in the width of the cortex rather than by an increase in intracortical porosity. Bending strength and energy absorption were significantly greater in the femora plated with the more flexible GFMM plate than the Vitallium plate.

Moyen et al. (1978) placed a thick cast chromium cobalt plate laterally on the intact right canine femur and a thinner, more flexible titanium alloy plate on the left femur and studied periodically the changes in the bone loss, bone mass, porosity, and remodeling. The plates were left on for a total of 6 months and in some animals bone remodeling was studied for an additional 3 months after plate removal. Total bone loss under the rigid plates significantly exceeded that under the flexible plates as measured by both total bone area and photon absorptiometry. A significant increase in intracortical porosity occurred under the rigid plate compared with under the flexible one. With rigid plates, the bone mass of the femur in the area beneath the removed plate was significantly greater than that on the contralateral femur where the plate was left in place for a total of 9 months. Following removal of the flexible plate, a slight increase in periosteal osteogenesis occurred, while after removal of the rigid plate there was a significant transient rise in osteogenesis intracortically, periosteally, and endosteally.

Pilliar et al. (1979) placed a porous surface coating, Co-base alloy on the surface of a 316L stainless steel AO plate. Each porous surface coated plate was screwed to the lateral cortex of the intact canine femur and a similar, uncoated plate was screwed to the lateral aspect of the contralateral femur. They observed bony ingrowth into the porous-coated plate during a period of 6 months. In both femora extensive bone remodeling and bone loss occurred. However, bone loss was more extensive in the femora under the plates with the intracortically porous surface layer. The authors attributed this greater bone loss to the more efficient stress transfer from the bone to the metal plate due, presumably, to better bone-to-implant bonding.

Bradley et al. (1979) studied bone remodeling under plates of a similar shape but with a wide range of elastic moduli. Plates were used that consisted of fiber polymer composites as well as metal plates. The plates were applied to the antero-lateral aspect of osteotomized canine femora and left in place for 16 weeks. The contralateral femur was used as a control. Bone specimens (composed of cortex and callus) were removed from the area beneath the plate and from the opposite cortex. The strengths of these bone material specimens were determined in bending tests. Bone tissue strength was inversely related to the rigidity of the plate used.

In the present study we used the results reported by Vasu et al. (1980) and Carter et al. (1981a) to estimate the *in vivo* stress fields in the normal canine femur during the mid-stance phase of the gait cycle. Determinations were then made of the alteration of these stresses due to the application of the various plates used in each of the previous studies described above. The results of these previous plating experiments were then interpreted in the light of the altered *in vivo* stress fields as calculated in this investigation.

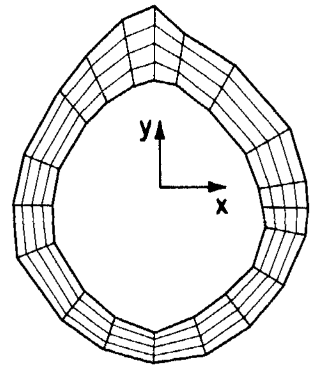

Figure 1. Element mesh of section of femur used for determining the bone sectional properties.

ANALYSIS

In a previous study reported by Vasu et al. (1980) and Carter et al. (1981a) *in vivo* strains were recorded from strain gages bonded at three circumferential sites at the femoral mid-diaphysis of a 35 kg mongrel dog. Strain recordings were made during gait at a velocity of 1.4 m/s. These strain measurements were analyzed with the use of mathematical models to estimate the axial load (P) and the bending moments M_x and M_y in the A-P and M-L planes during the entire gait cycles. The mathematical model was based on beam theory with a superimposed axial force (Rybicki et al. 1977). From the calculated force resultants during gait, the intracortical axial stress distributions could be calculated for the entire femoral cross section at any point in time using the combined flexural formula for axial loading and biaxial bending. These calculations showed that during the swing phase, the intracortical axial stresses were negligible. During the stance phase, a maximum compressive stress of –5.1 MPa was found at the posteromedial aspect and a maximum tensile stress of 2.3 MPa was established at the antero-lateral aspect. The neutral axis (line of zero stress) at this time in the stance phase ran from the lateral aspect to the antero-medial aspect.

The study noted above provides an excellent description of the cyclic load resultants and stress fields which are imposed on the femoral mid-shaft during gait. However, the dog which was used in the study was somewhat larger (35 kg) than those used in plating experiments by the other researchers (generally 20–30 kg). To investigate the influence of plating on the cyclic bone stress fields in these small animals, we cut a mid-femoral cross section from a 21 kg dog. This section was photographed and enlarged so that the geometry could be digitized. The enlarged section was divided into 80 discrete elements and the area (A) and moments of inertia (I_{xx}, I_{yy}, I_{xy}) were determined (Figure 1). We assumed that the cyclic stress fields imposed on this section were similar to those imposed on the section from the 35 kg dog. That is, 1) during the swing phase the intracortical stress fields were negligible, and 2) during the stance phase the stresses were –5.1 MPa posteromedially, 2.3 MPa anterolaterally, and 0.0 MPa laterally.

Using the assumptions described above, we calculated the resultant axial force and bending moments. During the swing phase, these resultants were negligible since no significant bone stresses were created. To calculate P, M_x, M_y during the stance phase, we employed the combined flexural formula

$$\sigma = -\frac{P}{A} + \frac{M_x I_{yy} + M_y I_{xy}}{I_{xx} I_{yy} - I^2_{xy}}\, y - \frac{M_y I_{xx} + M_x I_{xy}}{I_{xx} I_{yy} - I^2_{xy}}\, x \quad (1)$$

where

σ = axial stress at location (x,y)

A, I_{xx}, I_{yy}, I_{xy} = sectional properties

P, M_x, M_y = load resultants (unknown)

Since the axial stress (σ) was assumed at three different locations, the load resultants P, M_x, M_y were determined by solving the three simultaneous equations. The results of these calculations were that during stance the axial force (P) was equal to 127N (60 percent body weight) and that the bending moments in the A-P and L-M directions were 0.88Nm and 0.33Nm, respectively.

To determine the intracortical stress field during the stance phase for the unplated bone, the calculated force resultants were substituted back into Eq. (1). The axial stress could thus be determined at any (x,y) location in the section. The influence of plate attachment on the intracortical stress fields during the stance phase was determined using a variation of Eq. (1) which incorporates composite beam theory (Popov 1976). These calculations required that the location, geometry and elastic modulus of each plate under investigation be accurately modeled (Table 1). Each plate being studied was necessarily divided into elements of finite size analogous to the bone section shown in Figure 1. Calculations of the intracortical stress fields during the stance phase of gait were done on a PDP 11/03 minicomputer.

In our calculation of stress fields after plating we assumed that 1) the load resultants P, M_x, M_y were unchanged from those in the unplated femur, 2) there was an ideal connection between the bone and the plate (i.e. strains were continuous at the interface), 3) the bone tissue was homogeneous with an axial elastic modulus of 12.1 MPa (Carter et al. 1980) and 4) the bone was intact. Furthermore, no attempt was made to determine a superimposed static stress field which may have been introduced by the application of a plate with pretension.

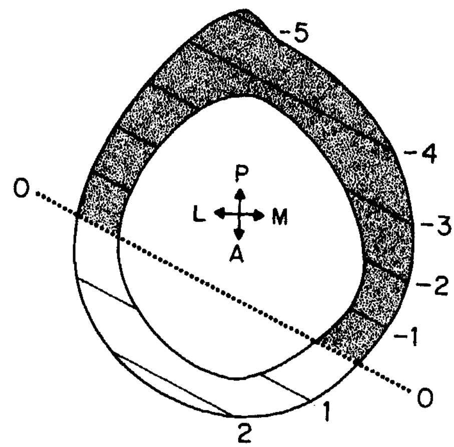

Figure 2. Stress distribution imposed on the unplated femoral mid-diaphysis during the stance phase of gait. Values are in MPa.

RESULTS

The intracortical axial stress distributions during the stance phase of gait for the unplated femur are shown in Figure 2. In the normal, intact bone, the maximum compressive stress during the stance phase is approximately –5.1 MPa on the posterior aspect and the maximum tensile stress is approximately 2.3 MPa at the antero-lateral aspect. Table 1 describes each of the plate configurations used by previous investigators which were modeled in this study. The intracortical

Table 1. Plating configurations analyzed

Figure Number	Reference	Osteotomy	Plate Pretension	Materials Modulus (GPa)	Plate Geometry (mm)	Bone Aspect
3	Uhthoff & Dubuc (1971)	Yes	Yes	Stainless steel (200)	10×4	Lateral
4A	Pilliar et al. (1979) Tonino et al. (1976)	No	No	Stainless steel (200)	11×3.8	Lateral
4B	Tonino et al. (1976)			PTFCE (1.2)		
5A	Akeson et al. (1976)	No	No	Vitallium (260)	11×3.8	Anterior
5B				GFMM (20)		
6A	Moyen et al. (1978)	No	No	Cast Ch-Co (208)	11×4.6	Lateral
6B				Titanium alloy (107)	11×2	
7A	Bradley et al. (1979)	Yes	No	Stainless steel (200)	11×3.8	Anterolateral
7B				Glass/Epoxy (65)		
7C				Graphite/Polysulfone (17)		

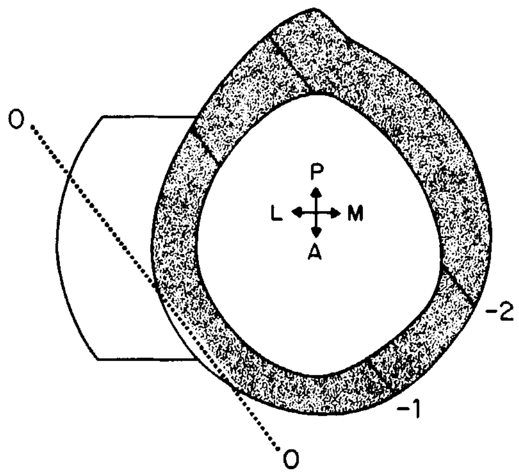

Figure 3. Bone stress distribution (MPa) during the stance phase of gait for the plate configuration used by Uhthoff & Dubuc (1971) (stainless steel plate).

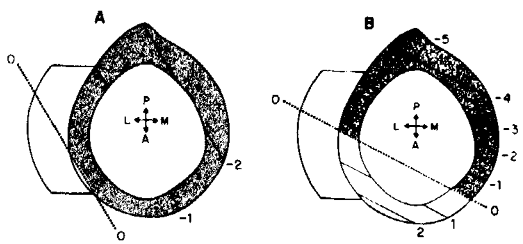

Figure 4. Bone stress distributions (MPa) during the stance phase of gait for the plate configurations used by A) Pilliar et al. (1979) and Tonino et al. (1976) (stainless steel plate) and B) Tonino et al. (1976) (PTCE plate).

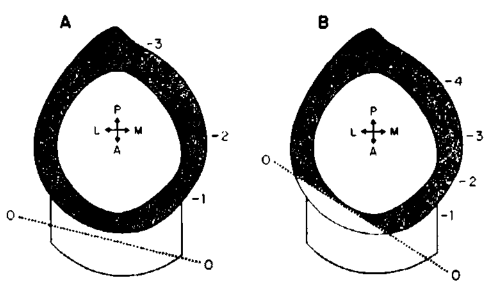

Figure 5. Bone stress distributions (MPa) during the stance phase of gait for the plate configurations used by Akeson et al. (1976) and Woo et al. (1976). A) Vitallium plate and B) GFMM plate.

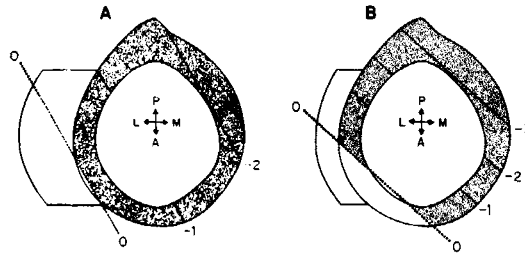

Figure 6. Bone stress distributions (MPa) during the stance phase of gait for the plate configurations used by Moyen et al. (1978). A) Cast Cr-Co plate, and B) Titanium alloy plate.

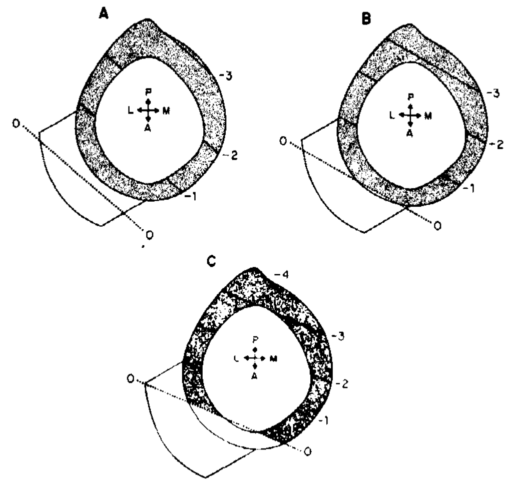

Figure 7. Bone stress distributions (MPa) during the stance phase of gait for three plate configurations used by Bradley et al. (1979). A) Stainless steel plate, B) Glass/ Eposy plate, and C) Graphite/Polysulfone plate.

stress fields calculated for the plate configurations used by these previous investigators are shown in Figures 3–7. In evaluating these stress fields, one should compare them to the stress distributions in the normal, intact bone (Figure 2).

The results of our calculations suggest that the canine femur is very sensitive to what may appear to be small changes in *in vivo* cyclic stresses. In the study reported by Akeson et al. (1976) and Woo et al. (1976), significant differences in bone remodeling and bone loss were observed when GFMM plated and Vitallium plated femora were compared. The cyclic bone stress fields calculated for these two plates are shown in Figure 5. Akeson et al. found that bone loss was less in the GFMM plated bone than in the Vitallium plated bone. The difference in bone mass of these two groups was attributed primarily to a greater degree of cortical thinning in the Vitallium plated bone (Table 2). The loss of cortical thickness in the Vitallium plated bone was greater in the anterior than in the posterior regions of the bone.

Moyen et al. found that Cr-Co plated femora had a greater bone loss than the titanium alloy plated femora (Table 3). The calculated cyclic stress fields for these two plating configurations are shown in Figure 6. The findings of Moyen et al. that bone loss was primarily caused by cortical thinning rather than a large increase in intracortical porosity, corroborated the earlier results of Akeson et al. In dogs plated for 9 months, the bone cross-sectional area was reduced 21 percent in the Ti alloy plated dogs and 31 percent in the Cr-Co plated dogs when compared to the control animals.

The morphometric findings of Akeson et al. and Moyen et al. are striking when viewed in the context of the cyclic stress fields calculated in the present study (Figures 5 and 6). In both of these previous studies, significant differences in bone loss was observed when a relatively stiff plate was compared with a more flexible plate. The maximum differences calculated for cyclic bone stresses, however, were in the range of 1 MPa for both of these studies. This difference in cyclic stress is less than 1 percent of the ultimate strength of the bone tissue and approximately 10 percent of the fatigue strength at 10^7 cycles (Carter et al. 1981b). This observation suggests that cortical bone remodeling is extremely sensitive to small changes in cyclic stresses.

*Table 2. Cortical thickness (mm) of GFMM and vitallium plated femora**

	LOCATION	
	Anterior (AL, A, AM)	Posterior (PL, P, PM)
9 Month Series		
GFMM Plated	2.3±0.3	2.3±0.4
	(not significant)	
Vitallium plated	2.0±0.4	2.3±0.2
	($P < 0.005$)	
12 Month Series		
GFMM plated	2.2±0.4	2.3±0.2
	(not significant)	
Vitallium plated	1.7±0.3	1.9±0.1
	($P < 0.05$)	

* Mean values and standard deviations from Akeson et al. (1976). Calculated cyclic stress fields shown in Figure 5.

*Table 3. Bone loss in chromium cobalt and titanium alloy plated femora**

	Total Area mm^2	Endosteal Perimeter (mm)
Controls	390	34.0
	(70)	(2.9)
6 Month Series		
Ti Plated	326	31.5
	(36)	(3.2)
Cr-Co Plated	287	33.0
	(50)	(3.0)
9 Month Series		
Ti Alloy Plated	308	30.5
	(27)	(3.6)
Cr-Co Plated	268	33.0
	(25)	(3.6)

* Mean values and standard deviations from Moyen et al. (1978). Calculated cyclic stress fields shown in Figure 6.

DISCUSSION

In this study we have calculated the mid-stance axial stress distribution in the normal dog femur and in femurs with plates of different stiffness. During gait, these stress fields are cyclic in that they are present only during the stance phase. In the swing phase of gait, the bone stresses are negligible. In relating the calculated stress fields in this study to the bone remodeling studies by other researchers, one must be aware that the cage activities of dogs do not consist of only walking in a straight line. Jumping, turning, lying down, and standing up will all impose different stresses on the bone than observed in normal gait. In addition, the remodeling response is probably sensitive not only to the cyclic stress distributions imposed but also to the number of loading cycles encountered. Nevertheless, the results of this study provide a general framework for assessing the change in cyclic *in vivo* bone stresses caused by the attachment of various fracture plates.

It was assumed that the strains at the bone plate interface are continuous. The study by Pilliar et al. (1979) casts some doubt on this assumption. Pilliar and associates found that a plate which allowed porous ingrowth at the bone interface resulted in greater bone resorption than a standard plate. They suggest that this greater resorption was due to a more efficient transfer of stress at the interface. This implies that our condition of continuous strains across the interface is better satisfied by a porous coated plate than a standard plate. With our model, we would calculate identical stress fields for these two plate types (Figure 4A). Further studies are needed to clarify the bone/plate interface mechanics. However, if strains are not continuous across the plate/bone interface, the true degrees of "stress shielding" created by plating would be less than that predicted by our calculations, not more. Our calculations represent the maximum stress shielding possible for that plate configuration.

Uhthoff & Dubuc (1971) applied a plate with pre-tension to the canine femur. Plate pretension creates a complex static stress field at the fracture site which will change with time (Askew et al. 1975, Perren et al. 1969). No attempt has been made in this study to account for this superimposed stress field.

Some general observations can be made about the influence of plate attachment on *in vivo* cyclic bone stresses. Plate attachment generally causes a shift in the location of the bone stress neutral axis (line of zero stress) toward the plate. The greater the shift in the neutral axis, the greater the change in the cyclic bone stresses. This amount of the neutral axis shift depends on 1) plate location, 2) the plate elastic modulus, and 3) the plate geometry.

The largest shift in the neutral axis would be caused by attaching the plate directly to the area of highest stress in the intact normal bone. This location is the posterior aspect of the canine femur which is a region of compressive stress. If one wishes to attach the plate to a neutral or tensile aspect, the largest shift in the neutral axis will be caused by applying the plate anterolaterally. This point is illustrated by comparing the results shown in Figure 4B (based on Tonino et al. 1976, and Pilliar et al. 1979) and 7A (based on Bradley et al. 1979). Both configurations used a standard AO stainless steel plate. However, when the plate was placed anterolaterally (Figure 7A) as opposed to laterally, there was a much greater shift in the neutral axis. In fact, the neutral axis moved off the bone cortex so that the entire cross section was exposed to compressive stresses.

The elastic modulus of the plate can have a strong influence on the amount of shift in the neutral axis. The greater the elastic modulus, the greater is the shift in the neutral axis and the "stress shielding" effect of the plate. This is illustrated by the results shown in Figure 4 (based on Tonino et al. 1976), Figure 5 (based on Akeson et al. 1976 and Woo et al. 1976), and Figure 7 (based on Bradley et al. 1979).

The influence of plate geometry on the shift of the neutral axis is similar to the effect of elastic modulus. Larger plates will cause a greater shift of the neutral axis than smaller plates of the same material. In the study by Moyen et al. (1978) a comparison of bone remodeling was made after plating with two plate types (Figure 6). The first plate studied (Figure 6A) was made of a stiffer material and was significantly thicker than the second plate studied (Figure 6B). Both of these characteristics tended to increase the "stress shielding" effect of the plate and caused a greater shift in the neutral axis.

In the mathematical models generated in this study, we did not attempt to simulate the osteotomy interface that was introduced by Uhthoff & Dubuc (1971) and Bradley et al. (1979). The characteristics of such an interface would allow the transmission of compressive stresses but not tensile stress. However, the results of our calculations show that with the metal plate used by Uhthoff and the stiffer plates used by Bradley et al., insignificant tensile stresses are created in the bone section. Since the stiff plates caused such a great shift in the neutral axis, nearly the entire bone section was exposed to cyclic compressive stresses (Figure 3, 4A, 4B). With the flexible plates used by Bradley et al. (Figure 4C), there was a region of bone tension at the anterolateral aspect. With an osteotomy, bone in this region would not be in tension but a small gap would be created (Carter & Vasu 1981). The calculated bone stress would also be slightly different from those shown.

CONCLUSIONS

The analytical methods used in this study provide a framework from which one can assess the influence of plate application on *in vivo* cyclic bone stress fields. The calculations that have been made suggest that bone remodeling is very sensitive to small changes in cyclic stresses. Future studies which carefully correlate regional changes in cyclic bone stresses to bone metabolic activity will hopefully elucidate the stress-remodeling relationships in plated long bones.

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