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AN ELASTIC ANALYSIS OF A PLATED BONE TO DETERMINE FRACTURE GAP MOTION

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SUMMARY

An elastic analysis to determine fracture gap motions occuring in the osteotomized and plated canine femur was performed using the finite element program NASTRAN. The femur was idealized as a hollow right cylinder, and transverse anisotropy was assumed for the elastic properties of the bone. A 3-D 360 degree model consisting of 224 isoparametric quadrilateral hexahedral and 11 beam elements was created. A range of plate stiffnesses was tested by varying the modulus of elasticity of the plate from 207 GPa to 1 GPA. Moments were applied in the plane of the plate, about the axis of the plate, and in the plane of the screws.

Results showed that, for plates of typical geometry and elastic modulus under 10 GPa, the contribution to fracture gap motion occuring due to deformation in the bone was negligible compared to that contribution from deformation in the plate. Fracture gap motion for bone-plate systems using plates in this modulus range could therefore be estimated using beam theory. Deformations occurring in the bone were relatively unaffected by plate stiffness. The relationship of gap motion to plate stiffness was non-linear above 10 GPa due to the decreasing contribution of the plate's deformation.

INTRODUCTION

The use of a compression plate as a method of treating severe bone fractures is a currently accepted practice in orthopedics. In this design (Figure 1), the plate is applied in such a manner that it locks in a small amount of press-fit between the fragments; the applied force and close fit of the fragments induces the bone to grow faster. However, the demand for this close, solid fit means the compression plate must be stiff - so stiff, in fact, that the plate continues to carry much of the load on the bone even after the fracture has healed. Most bones need strains imposed on them to maintain their size and strength; in the absence of these strains, they atrophy in a manner similar to unused muscles. It has been shown experimentally (8,9,10,11) that the application of a stiff bone plate to an intact bone will generally lead to a long-term loss in the volume of bone under the center of the plate. The resulting decreased wall thickness under the plate midspan makes the bone prone to refracture after plate removal.

To solve the problem of stress shielding, a number of researchers have experimented with an increasingly more flexible series of plates. The most obvious lower bound to plate flexibility occurs when the plate permits so much motion of the fragments that it is no longer possible for the fracture gap to be bridged. As the plate stiffness approaches the non-union point, the time required for the fracture to heal increases to infinity. In much of this research, the osteotomized canine femur has been used as a model. Selected results of these tests, in terms of elapsed time before union of the fracture versus plate stiffness, are shown in Table I.

We were interested in the possibility of a resorbable plate but, since the resorbable material we were working with was fairly flexible, we were unsure whether we could make a plate stiff enough to be above the non-union bound. While estimates of the value of this bound could be made from the data of other researchers, we wished to define it as accurately as possible before making our own plates. Before starting a series of implantation studies in dogs to define this bound, we performed a mechanics analysis to determine the characteristics of the bone-plate system. This analysis is the subject of this report.

METHODS

It was assumed that motion at the fracture gap was the governing variable determining whether or not a fixed fracture would go to union. To determine the total deflection at the gap, it was necessary to determine how much of the total deflection was due to plate deformation and how much was due to bone deformation. While it was easy to estimate the plate bending using beam theory, defining the bone deformation required a complex mechanics analysis easiest done, and perhaps only possible, with the finite element method. It was expected that the application of loads from the relatively narrow plate to the bone might cause the bone to deflect significantly in the area of the plate. This was especially true because our plate design created a line contact between the plate and bone. Since this effect could not be simulated with a two dimensional (2D) model, it was decided to create a three dimensional (3D) model with the bone composed of solid quadrilateral elements, and the plate and screws of one dimensional (1D) beam elements.

The dog femur was modelled as a hollow right cylinder. The outer radius of 9.5 mm and the inner radius of 6.9 mm were determined from inspection of available femurs and the published data of other investigators, and are intended to represent dogs in the 18 to 27 kg range. The length of bone modelled was chosen to be the length covered by the plate plus one bone diameter more past the end of the plate to allow point loads applied at the end to spread out.

The geometry of the plate (Figure 2) had been previously fixed as a preliminary to the manufacture of prototype plates. It was intended that factors such as orientation of the composite layers and percentages of resin would be varied to produce the desired stiffneses. This effect was simulated in the model by simply varying the modulus of elasticity of the beam elements. The high stiffness plate used the elastic modulus of 316L stainless steel. The geometry and size of our plate was approximately what would be chosen to fix a bone this size in clinical practice. The results for the high stiffness plate could therefore be considered to represent a typical case where stress shielding might be a problem. The modulus used for the medium stiffness plate was chosen to produce a stiffness slightly above the estimated non-union bound. The modulus used for the low stiffness plate produced a stiffness below the non-union bound.

Values for the elastic modulus of canine bone were available in the literature, but were taken under conditions inappropriate for the loadings anticipated in this study. The material properties used were those determined for human cortical bone by Reilly and Burstein (7); these properties assume transverse anisotropy. The bone was idealized as being composed of purely cortical bone, no attempt was made to model cancellous bone.

With these dimensions settled, a model was then created using the NASTRAN finite element program. The input deck to NASTRAN was generated using the PATRAN graphics pre- and post-processor. The model was made to be roughly the maximum size that would allow reasonable execution times on the DEC VAX-11/780 computer. The model consisted of 224 solid isoparametric quadratic CIHEX2 elements and 11 CBAR beam elements, and executed in eight hours cpu time. A smaller, faster executing 180 degree model was also constructed for use in those cases where it was possible to assume symmetry about the plane of the screws(Figure 3). It was not possible to run a more finely divided model to test for convergence. The results shown in this report will be checked against a similar physical model in an experimental program currently underway. Additionally, a model similar to the 180 degree model with nearly identical mesh but isotropic material properties was tested against a strain-gauged test model by Cheal et al. (5) and shown to be accurate.

In order to most efficiently utilize the available computer time, only those cases which would produce the most significant deflections were run. Axial compression was not applied since it was felt that, because loads were applied to the axis of the plate, this load case would result in only minor deflection in the plate and a small shear deflection in the bone. A force causing a moment in the plane of the screws, directed so as to open the osteotomy, was applied on the axis of the plate at the far end of the bone. This is referred to here as the bending open or BO load case. A moment was applied about the axis of the plate; this axial torque is referred to as the TA load case. A force causing a moment in the plane normal to the axis of the screws was applied on the axis of the plate at the far end of the bone. This moment, which produced a bending against the plate's stiffest direction, is referred to here as the BA load case.

In the final "load case", an axial torque was again applied, and also a configuration change was made. The plate-bone contact was changed from a line contact at the axis of the plate to a two line contact. These lines were at the axis of the plate and the outer edge of the plate which would be driven into contact with the bone by an axial torque. This load/configuration case, which modelled a "wide-wheelbase" plate, is referred to as the WW load case.

RESULTS

The fracture gap motions resulting from the various load cases are presented in Table II. The deformed shapes are presented in Figures 4a-d.

DISCUSSION

The highest deflections were produced by the BO and TA load cases. The BA load case produced deflections which were almost negligibly small in comparison.

For the BO load case, the deflection took place primarily in the plate. This was especially true for the lower stiffness plates. With the high stiffness plate, significant deflection (accounting for 34% of the motion at the gap) occurred in the bone. This was not a whole-scale bending of the bone, but instead appeared to be a local deformation under the plate. This was especially visible as a raised dimple under the screw closest to the fracture. A shallow depression under the far end of the plate was also visible. This would seem to indicate that the area under the screws (where load transfer occurs) was highly stressed, consistent with the proliferation of bone seen in this area on X-rays. A possible hypothesis is that what really affects fracture gap motion is not the bone's moment of inertia as a cylinder, but the stiffness of the wall under the plate. However, this is not a critical point since, for the low stiffness plates, the motion at the gap was almost entirely a function of the plate stiffness.

Deformation under axial torsion (TA) also took place primarily in two modes. The first mode, responsible for the great majority of the motion at the gap, was twist of the plate in the span between the fracture gap and the closest screw. The second mode was twist of the bone cylinder itself. This was less visible on the plots with flexible plates because the twist of the plate was greatly increased, making the bone twist appear relatively small (the deflections for plots were scaled by the computer for readibility). The torsional twist of the bone appears to be an action of the whole bone, rather than a local action as was the case under the BO load.

The BO and TA load cases produced roughly the same amount of deflection at the fracture gap for equivalent applied moments.

Bending in the plane of the plate took place mainly as a bending of the plate. The beam element bent against their stiffest direction. The fracture gap motion produced was more than an order of magnitude less than the deflections produced by either the BO or TA load cases, on an equivalent applied moment basis.

Changing the plate bone contact to the wide wheelbase configuration and applying an axial torsional load produced a noticeable flattening of the bone cylinder under the outer plate edge. The bone appeared to be more distorted in this case than in any other, yet fracture gap motion was only decreased by one-third. This leads to the conclusion that the wide wheelbase plate design may not be necessary if the screw/plate fit can be relied upon to provide rotational stiffness.

The fracture gap motions predicted are quite high compared to what might be expected to be the maximum movement allowable for healing. Carter et al. (4), implanted strain gauges on the femur of a 35 kg dog and studied the strains under normal gait. From these strains, they calculated the loads at the instant of maximum net loading of the mid-femoral diaphysis as a 250 N axial compression, a 2 5 N-m bending moment directed to produce tension anteriorly, a 1.4 N-m bending moment producing tension laterally, and a torsional moment of 0.56 N-m where the vector points proximally. These loads produce hypothetical fracture gap motions of 52 to 0.33 mm due to bending moments, and 9.0 to 0.44 mm due to torsional moments, depending on the stiffness of the plate used. With the medium stiffness plate, the predicted fracture gap deflections are 5.0 and 1.1 mm, for bending and torsional loads respectively. A possible hypothesis is that these magnitudes are realistic, but that the deflections in bending stay low because the plate is positioned so that it is usually in tension (i.e., bending to open the osteotomy doesn't occur), and that the torsional displacements do occur and are the relevant concern in the design of fixation plates. In cases of fractures, the interdigitation of the ragged fracture edges could be expected to lend support and there may be some stiffening due to the surrounding tissues; neither of which was considered in this model. The main conclusion from this result though is that unusual care probably would have to be taken of a fracture stabilized with a flexible plate.

The total deflection of the system was a product of the deflections of the plate, the screws in the bone, and the bone itself. In this analysis, only the plate stiffness was varied, and therefore the deflection in the bone and screws was relatively constant (although affected somewhat by the bracing action of the plate). The two ends of the deflection-stiffness curve, that is the response at very high and very low plate stiffnesses, could then be estimated. In the case of very flexible plates, the plate would deform so much that the deformation of the bone and screws would be negligble by comparison. In this case, the deflection would vary with the inverse of the stiffness and have a constant slope which could be calculated by beam theory. In the case of an infinitly stiff plate, the deflections present would all occur in the bone and screws. Therefore, in this case the deflection would be constant; that is, it would not vary (or would vary only a negligible amount) as the beam stiffness was varied. Using this reasoning to predict the shape of the tails, the general shape of the deflection curve can be predicted (Figure 5).

The calculated results show that the deflections varied in nearly exact proportion with the plate flexibility across the range of the two lower stiffness plates tested. Even as the plate stiffness greatly increased to the stiffness of a typical solid stainless steel plate, the bending in the bone contributed only a small portion of the fracture gap motion. This implies that, for flexible plates, the fracture gap motion is a function solely of the plate stiffness and can be directly calculated from beam theory.

TABLE I. Materials evaluated using the canine femur model

Investigator	Material	EI (Nm2)	Results (weeks)
Bradley (1)	SS	16.80	U
	GEC	5.48	U
	GPC	1.45	U
Brown (2)	PA	0.49	4
	PBT	0.41	4-5
	PBT	0.23	6
	PBT	0.49	12-13
Brown (3)	PA	0.265	4-12
	PBT	0.221	4-6
	PP	0.239	22
	UHMWPE	0.161	>40
Zenker (11)	GFMM	3.01	U
Kusenose (6)	M90	0.040	9-30
	CR20	0.067	38->40
NASTRAN model	high	6.02	-
	medium	0.29	-
	low	0.029	-

Legend:

GFMM - graphite fiber reinforced methyl methacrylate
SS - stainless steel
GEC - glass-epoxy resin composite
GPC - graphite reinforced polysulfone
PA - polyacetal
PBT - polybutyleneteraphtalate
PP - polypropylene
UHMWPE - ultrahigh molecular weight polyetylene
U - proceeded to union, no exact time given

TABLE II. Calculated fracture gap motions

Load case	Plate stiffness	Slope (1/Nm)	Fracture gap (mm/Nm)
BO	high	0.00304	0.116
BO	medium	0.0458	1.74
BO	low	0.474	18.0
TA	high	0.00659	0.25
TA	medium	0.0506	1.92
TA	low	0.424	16.1
WW	high	0.00440	0.167
BA	low	0.0489	0.929

BO - bending to open the osteotomy

TA - axial torsion

WW - axial torsion, wide wheelbase plate

BA - bending in the plane of the plate

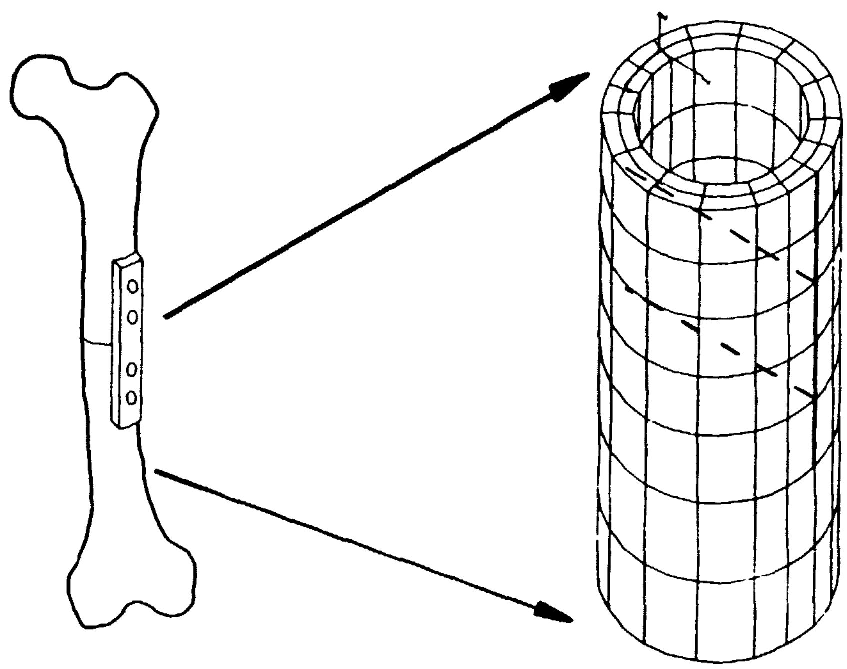


Figure 1. Derivation of model

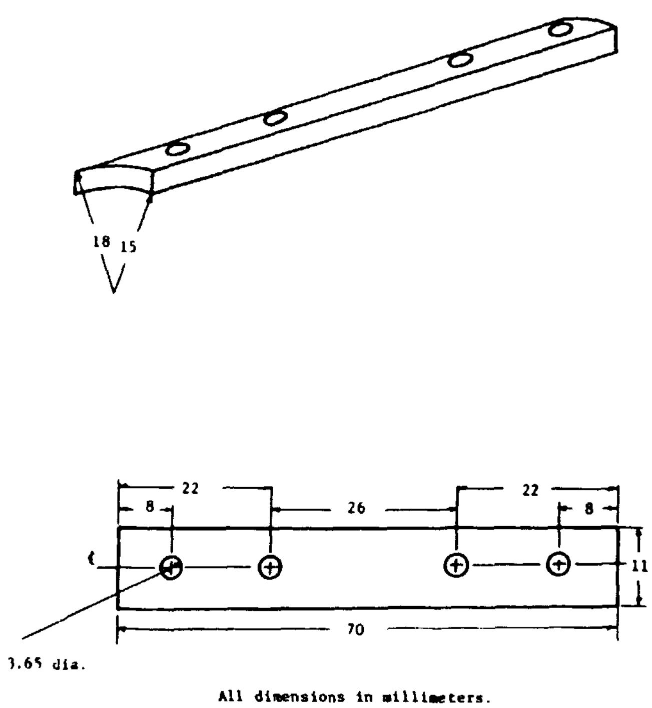
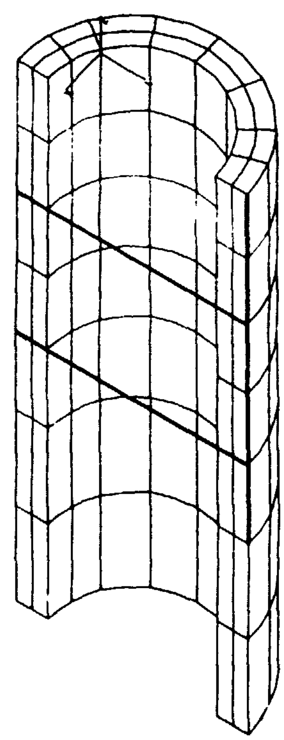
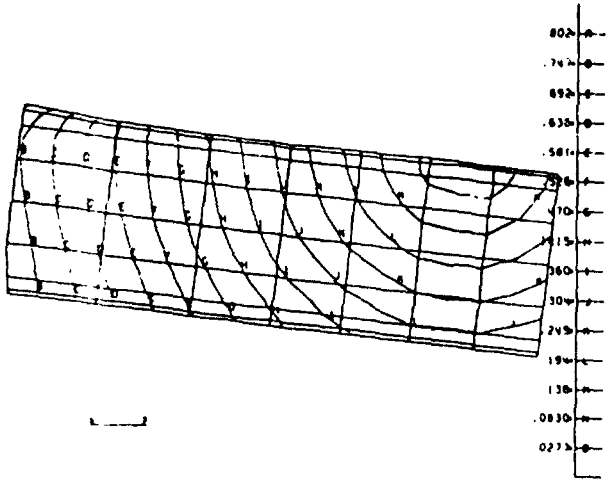


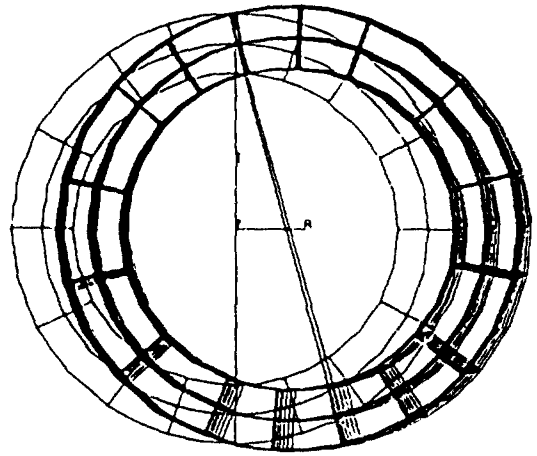
Figure 2. Plate geometry

Figure 3. 180 degree model



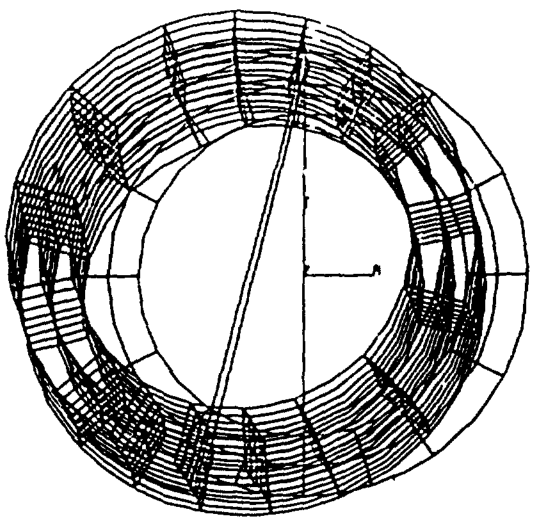


BO load case

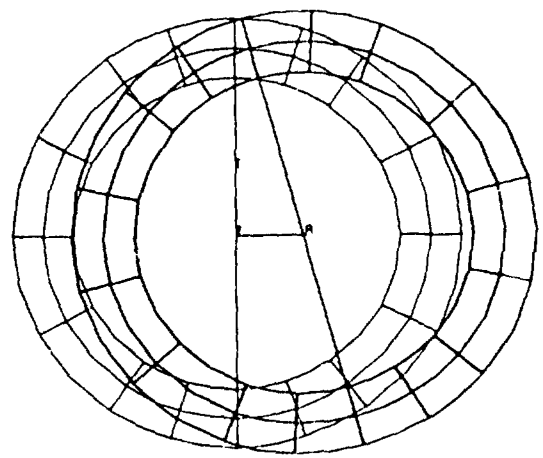


TA load case, high stiffness

WW load case



TA load case, low stiffness



Figures 4a-d. Deformed shapes

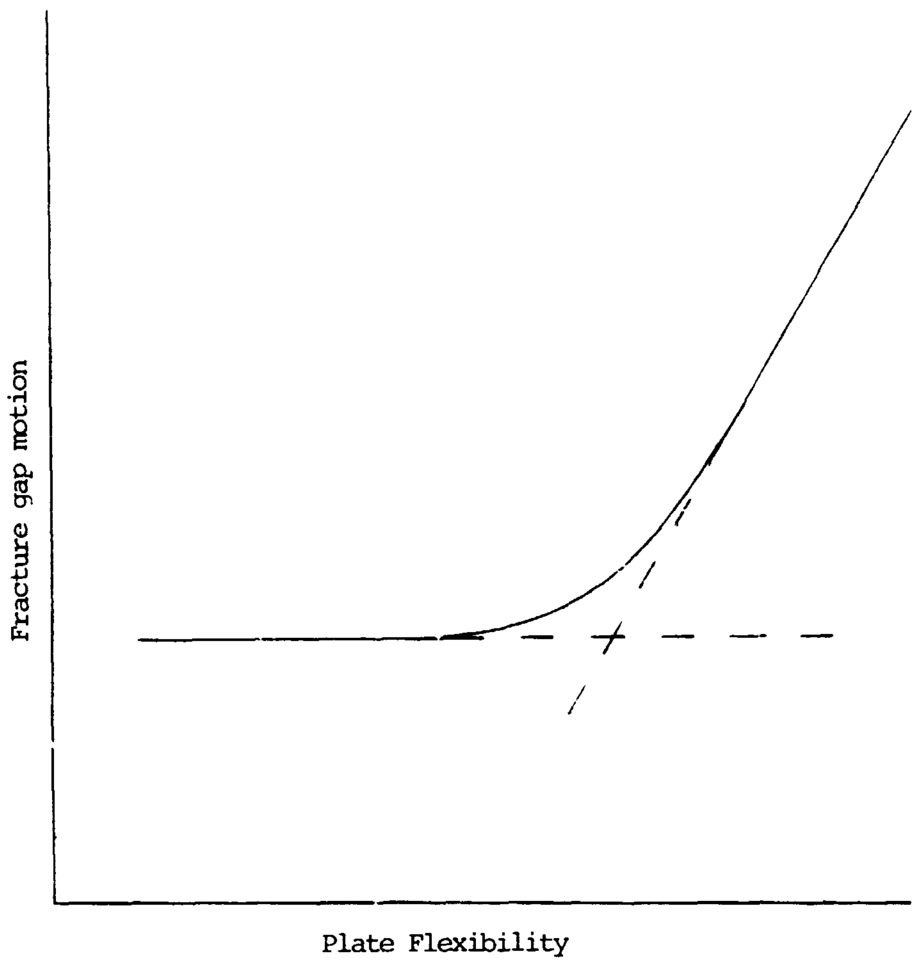


Figure 5. Predicted fracture gap motion

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