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FINITE ELEMENT DESIGN OF A MECHANICAL TESTING METHOD
FOR POLYMER COMPOSITE FEMORAL STEMS

by

ANNELIESE DOROTHY HEINER

Submitted in partial fulfillment of the requirements
for the degree of Doctor of Philosophy


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January 1995
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FINITE ELEMENT DESIGN OF A MECHANICAL TESTING METHOD FOR POLYMER COMPOSITE FEMORAL STEMS

Abstract

by

ANNELIESE DOROTHY HEINER

Polymer composite femoral stems do not have a standardized in vitro mechanical testing method. The testing device for metallic stems, which simulates proximal resorption of the femur, is inappropriate for composite stems, which they are expected to reduce or prevent proximal resorption. The objective of this study was to determine a mechanical testing method for pressfit composite stems, using finite element analysis (FEA). The goal was to reproduce the strains of the stem implanted in a femur.

Three simplified models used for preliminary studies were beam-on-elastic-foundation, two-dimensional plus sideplates FEA, and axisymmetric FEA. These models showed that stem stresses were affected by design changes to the testing device, and gave some results equivalent to those found with the more complex models.

The effects on the composite stem of varying the testing device were studied using full 3-D FEA models. The stem strains were affected by design changes to the testing device. The maximum normal and interlaminar shear strains of the composite stem in the testing device were either lower than or about equal to the maximum strains of the composite stem in the femur.
A single-material testing device made of birchwood was chosen as the best testing device from the parametric study. Birchwood is an orthotropic material with a longitudinal stiffness in the range of bone. This testing device improved the stem strains which lead to composite interlaminar failure, relative to the other testing devices studied.

The viability of the testing device, as measured by stresses on the bone cement or testing device and by interface motion, was also affected by changes to the testing device and stem material. Changing from a metallic stem to a composite stem increased the bone cement stresses by about twofold. Some testing device configurations had cement stresses that were close to or above static cement strengths. The birchwood testing device had high interface motion.

Other stem materials were compared to the composite stem. The strains on a metallic stem and the composite stem were not always affected in the same way by changes to the testing device. An isotropic stem was a better simplification of the composite stem than was a transversely isotropic stem.
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1. Introduction

Femoral stems which are made of a rigid material or are cemented into the femur can cause stress shielding and proximal resorption of the femur. This proximal resorption is modelled during mechanical testing by fixating only the distal part of the stem, and loading the top of the stem. Polymer composite femoral stems, and stems which are bonded to the proximal part of the femur, should reduce or prevent proximal resorption. Because of this, a mechanical testing method for femoral stems designed to prevent proximal resorption, such as composite stems, should be different from a mechanical testing method for femoral stems which cause proximal resorption.

1.1. Mechanical Testing Configurations for Femoral Stems

The most commonly used mechanical testing configuration for femoral stems is distal potting, a configuration in which only the distal portion of the stem is supported and a load is applied to the head (Fig. 1.1.a). This testing configuration models proximal resorption of the femur, as would be seen with metallic stems or with stems which have been cemented into the femur. For metallic stems, this testing configuration produces failures that are similar to in vivo failures [1]. Distal potting is found in national [2-4] and international [5,6] standards. A variation of this configuration had the head fixed and the distal tip of the stem loaded [7]. Distal potting has been used to study the effects on femoral stems with changes in the angle of the applied load, cyclic load range and...
frequency, level of embedding, and the material and geometry of the stem [1,7,8-13].

Two other testing configurations studied for metallic stems are three-point bending (Fig. 1.1.b) and four-point bending (Fig. 1.1.c). The stresses on an implanted femoral stem depend on the bone cement, bone, and location of the neutral axis of the stem-cement-bone system [14]. Likewise, the stresses on a femoral stem potted in a testing device depend on the embedding compound, fixture, and location of the neutral axis of the system. The three-point and four-point bending configurations remove the influence of geometric and material variables of a testing device into which a stem would be potted. They claim easier reproducibility of the testing setup, since the load angle is easier to reproduce, there is no embedding height to reproduce, and there is no embedding compound to degrade and affect the stem stresses [14,15]. Three-point bending configurations have compared the mechanical properties of metallic stems with different materials and designs [14,16,17]. A four-point bending configuration tested the mechanical properties of one type of metallic stem [15].

A testing configuration referred to as proximal potting models the case when no proximal resorption of the femur occurs; the femoral stem is supported up to its neck (Fig. 1.1.d). A variation on this testing configuration had proximal and distal stem support; the middle portion of the stem was free [18-20]. Another variation was to create a proximal medial gap of up to 3 mm by rocking a proximally potted implant back and forth before the bone cement had completely
hardened, to model intraoperative loosening [7]. Still another variation on proximal potting was to include a small medial gap between the stem and the support, to model slight proximal resorption [21]. Polymer composite femoral stems, or stems with proximal porous coating, may reduce or prevent proximal resorption. Proximal potting is found in the literature [7, 8, 21-26], international standards ISO 7206/5 [27] and 7206/6 [28], and an ASTM standard draft [29]. However, because of the newness of composite stems, there is no large collection of in vivo failed stems with which to compare stems which have been failed using this testing configuration. Additionally, the details of the proximal potting, such as geometry and dimensions of the configuration, have not been standardized. This suggests that this configuration, and variations on this configuration, should be evaluated.

For proximal or distal potting of a femoral stem, the testing device is usually a metal fixture and an embedding compound. The embedding compound is usually bone cement [4]. Epoxy casting resin and high-alumina cement have also been suggested as embedding compounds [6]. Carbon fiber reinforced polyurethane has been used as a low-stiffness embedding compound for the testing of composite stems [23, 26]. Carbon fiber reinforced polyethylene has been proposed as a fixture having a stiffness similar to that of the femur [30]. Distal-potted testing devices of a single material have been made out of acrylic [7], epoxy [12], and metal [31]. The dimensions of femoral stem testing devices in the literature were rarely stated.
The diameter of the acrylic testing device just mentioned was 6 cm; the authors considered the stem to be rigidly fixed [7]. The inner diameter of the metal fixture in another testing device was 3 inches [26]. The ASTM standard for fatigue testing of metallic stems, F1440-92, recommends a cement thickness of at least 10 mm [4]. It has been recognized that the stem stresses may be affected by variables of the testing device, and that the stem stresses would be different when the stem is implanted in an actual femur [7]. However, a testing device allows for parametric studies to be done on a stem.

The load angle at which the femoral stem is tested simulates differences in surgical positioning (such as varus-valgus) [7,8,12] as well as differences in load angles during phases of gait and other activities. The load angle affects the stem stresses; torsional or out-of-plane loading, especially, is important for getting realistic failures in metallic stems [9-11]. Torsional loading is seen as a problem for uncemented stems, since torsional loading can cause loosening [32,33]. Since composite stems are intended to be uncemented, torsional loading should be included in a load angle for testing composite stems. The exact testing angles are also a factor; too much torsional loading (too high of an out-of-plane angle) was found to be too severe a loading condition for metallic stems [9].

Different embedding heights simulate different degrees of loosening or proximal resorption [7,8]. The simulation is not exact, since a proximally resorbed femur could still provide some support to
the stem [7]. The embedding height can have a profound effect on the stem stresses [7], and is a variation used in the testing of metallic stems to obtain realistic stem failures [1].

1.2. Polymer Composite Femoral Stems

1.2.1. Polymer Composites

A composite material is a combination of a discontinuous material and a continuous material [34]. The composite retains some of the characteristics of the original materials, resulting in improved characteristics of the composite over one of the original materials. Polymer composites combine a polymer matrix with a filler or a reinforcement such as carbon fiber. One type of composite used for orthopaedic and prosthetic devices is a layup, an assembly of layers, of continuous-fiber reinforced polymer. These layers, called plies, can be oriented at different fiber angles, and can be assembled in different orders or stacking sequences. Varying the matrix, reinforcement, ply orientation, stacking sequence, or processing method varies the mechanical properties of the final composite. This is a major advantage of polymer composites over metals in the area of orthopaedic and prosthetic devices. Most importantly, polymer composites can be designed to be much less rigid than the metals used for these devices. This lower rigidity may reduce or prevent stress shielding of the bone [35]. Other advantages of polymer composites over metals are that the fatigue strength of polymer composites can be higher than that of stainless steel, even when the ultimate strengths of the two materials are
similar [36], they are invisible to x-rays [37], and they have better
corrosion resistance than metals [37].

1.2.2. Polymer Composite vs. Metallic Femoral Stems

Three differences between polymer composite femoral stems and
metallic femoral stems are stiffness, interface with the bone, and
anisotropy of the material. These differences suggest that a testing
configuration which is appropriate for metallic stems may not be
appropriate for composite stems.

Polymer composites can be designed to have a lower stiffness
than the metals used for prosthetics. The lower stiffness of
composite stems may reduce or prevent proximal resorption in the
femur, by reducing stress shielding of the femur. Testing devices
for metallic stems model proximal resorption by only supporting the
distal portion of the stem; this configuration might not be
appropriate for composite stems which do not cause proximal
resorption to occur. Additionally, even if some proximal resorption
did occur with a composite stem, the lower stiffness of the composite
may allow the stem to displace far enough under load so that it will
contact and be supported by the remaining proximal bone [38].

Composite stems interface with the bone through a pressfit or
through bonding to the bone, while metallic stems interface with the
bone through cement or through bonding to the bone. Different
interfaces between the stem and the bone will cause different loads
to be seen on the stem and the bone.
The types of polymer composites used for femoral stems are anisotropic, while the metals are isotropic. The anisotropy of composite stems changes the stresses that develop, and possibly changes the in vitro simulation of these stresses. A simpler testing device might work for evaluating metallic stems, but be too simple to adequately evaluate composite stems.

1.2.3. Previous Studies of Composite Stems

For composite stems, the matrices investigated have been polysulfone [39-45], polyaryletherketones such as polyetheretherketone (PEEK) [46,47], epoxy [40,48], pyrolytic carbon [49], and polyethylene-hydroxyapatite [50]. The reinforcement used is continuous carbon fibers. The composite layup has been studied for its effects on stem stiffness and strength [40], and for optimizing load transfer to bone [51,52]. The composite layup and stacking sequence are usually proprietary and not reported.

Mechanical tests of composite stems have shown satisfactory results. The reported static strengths of composite stems are comparable or superior to those of metallic stems [44,45,49]. Although the fatigue strength of composite stems is a concern [43,53], it has not yet been shown to be a problem [41,43,45,49].

Composite stems have been effective in canine studies. Maçee et al. implanted smooth, pressfit composite stems and found that proximal resorption did not occur [39]. Bone quality was maintained in the proximal femur at 6.5 years postoperatively [39,54-57]. The average proximal medial cortical thickness increased by 30% over that
of the unimplanted femur [57]. The neat polymer coating had cracks and surface indentations caused by wear [56]. The molecular weight of the polymer decreased slightly but unimodally, indicating that no structural breakdown had occurred. No degradation of the polymer was found using H-1 NMR. Cheal et al. found that pressfit composite stems resulted in no significant difference in bone remodelling after 2 years postoperatively, when compared to titanium stems [46,58]. Composite stems with a spongy polymer porous surface, investigated by Mendez et al., showed good results in fatigue, bone anchorage and ingrowth, and pullout strength at 8 months postoperatively [41,45]. Sumner et al. compared titanium and composite stems, both with a titanium fiber porous coating, and found that the femurs with composite stems had less loss of proximal cortical bone at 6 months postoperatively [47].

Christel et al. provide the only reported clinical trial of composite stems [49]. The composite stem was carbon-carbon. This trial was discontinued after 2 years and development of this stem stopped because of material problems and high development costs [46,59].

1.2.4. Mechanical Testing of Composite Stems

There has been considerable variation in the mechanical testing of composite stems. The international standard ISO 7206/6 for testing the head and neck of femoral stems [28], and an ASTM standard draft for specifically testing composite stems [29], had the stem proximally potted. The recommended applied load angles and
magnitudes were different. Specific testing device configurations were not given. Christel et al. mechanically tested a carbon-carbon stem using a test method based on an ISO draft involving distal potting [49]. Ainsworth and Tarr designed a composite stem based on the stresses produced using distal potting [44]. A mechanical testing device developed by Humphrey and Gilbertson and designed specifically for a composite stem was the variation on proximal potting mentioned earlier, where the proximal and distal femoral stem were supported and the midstem was free [19]. Leaving the midstem free was based on the assumption that the midstem supported much less load than did the proximal and distal stem. The response of a composite stem in this testing device was compared to a composite stem either pressfit or cemented (to represent a well-fixed implant) into an artificial composite femur. The testing device gave more support to the femoral stem than did the pressfit case, but less support than did the cemented case; there was better correlation with the cemented case. However, the authors noted that it is probably more desirable to more closely model the pressfit case, since a pressfit implant better represents a worst-case support condition than does a well-fixed implant.

1.3. Objectives

The main objective was to study the design of a mechanical testing method for a pressfit polymer composite femoral stem, using finite element analysis. The goal was to determine a mechanical testing method which came closest to reproducing the stem strains on
a finite element model of a composite stem implanted in a human femur. Other objectives were to investigate how testing device variables affected the viability of the testing device, to investigate the effects of different stem materials, and to demonstrate that a testing method that modelled no proximal resorption was different from a testing method that modelled proximal resorption.

Preliminary design studies were first done using beam-on-elastic-foundation theory, two-dimensional plus sideplates finite element models, and axisymmetric finite element models. These simpler models were used to evaluate the effects of design changes to a testing device and femoral stem, and to help determine what design variables to examine further. Next, a full, three-dimensional finite element model was used to parametrically determine how design changes to the testing device affected the composite stem and testing device, and to compare the stem strains to those of a stem implanted in a human femur finite element model.
Fig. 1.1. Mechanical testing configurations for femoral stems.

a. Distal potting
b. Three-point bending
c. Four-point bending
d. Proximal potting
2. Beam-on-Elastic-Foundation

2.1. Introduction

Preliminary studies of an implanted femoral stem were done using beam-on-elastic-foundation (BEF) theory. BEF theory gives equations for the stress state of a beam which is on an elastic foundation [60]. Here, the beam represented the femoral stem, and the elastic foundation represented the femur or testing device into which the femoral stem was inserted. Previous investigations have determined that general trends for the consequences of design changes related to intramedullary devices implanted in the femur are predicted well by BEF theory [61,62]. Additionally, BEF theory is more suitable for parametric analysis when compared to numerical methods such as finite element analysis, since the BEF models give mathematical formulas, in which the effects of parameters can be more quickly and easily determined [63]. In this study, BEF theory was used to show that the stresses on a simple femoral stem model depended on parameters of both the femoral stem and the testing device.

2.1.1. Theory

Beam-on-elastic-foundation equations combine the differential equation of a beam in bending, \( M = - EI \left( \frac{d^2y}{dx^2} \right) \), with the assumption that the reaction force, \( p \), in the elastic foundation is proportional to the deflection of the beam, \( y \) [60]. The constant of proportionality, \( k \), is the foundation modulus, so \( p = ky \). The solution of the resulting differential equation is integrated to
obtain $y(x)$. Within a continuous section of the beam (where $y$ and all derivatives of $y$ are continuous), the integration constants are determined through the boundary conditions at the ends the beam. Slope, moment, and vertical shear on the beam are calculated from the equation for $y$, as indicated below:

\begin{align*}
\text{Slope} &= \theta = \frac{dy}{dx} \\
\text{Moment} &= M = -EI \left(\frac{d^2y}{dx^2}\right) \\
\text{Vertical Shear} &= Qv = -EI \left(\frac{d^3y}{dx^3}\right)
\end{align*}

The foundation modulus, $k$, is related to the elastic modulus of the foundation, $E_f$, through the equation:

\[ k = (E_f) \cdot \text{(width of foundation) / (depth of foundation)} \]

2.1.2. Literature

Use of BEF equations to study implanted femoral stems was first reported by Huiskes in 1979 [64]. Huiskes developed approximate equations which calculated the most important stresses in the proximal, middle, and distal regions of the bone-cement-stem system. These equations were used to evaluate the influence of parameters such as stem length, modulus, and thickness, cement modulus and thickness, and bone quality and dimensions [61,64,65]. Guidelines for femoral stem designs and implantation procedures were developed based on these results. For instance, a minimal stem length could be
determined when stem thickness, stem modulus and geometric bone properties were chosen; using titanium instead of cobalt-chrome resulted in a 25% shorter stem length [61]. Proximal cement stresses could be minimized by using a stem thickness of about 85% of the width of the medullary canal [61]. The simplified BEF models of the bone-cement-stem system were comparable in some aspects to finite element models [61,64].

In a parametric analysis of pin-bone stresses in external fracture fixators, Huiskes et al. used BEF theory to predict the most important stresses in the pin-bone system [63]. These predictions compared well with finite element analysis predictions. The BEF model indicated that the pin-bone interface stress could be lowered by using a bilateral instead of a unilateral frame, decreasing the distance between the bone and the sidebar, increasing the pin elastic modulus, or increasing the pin diameter.

De Beus et al. developed a double-layer BEF model of an implanted femoral stem system which included a metal shell within the cement [66]. The purpose of the shell was to reinforce the cement and make revision surgery easier. Variables studied were stem length, shell length, stem radius, shell inner and outer radii, and bone inner radius. The system was optimized for the smoothest load transfer between the parts.

Bechtold and Riley used BEF theory for parametric analysis of intramedullary rods [62]. The theory was refined to take into account the nonconstant foundation modulus of the femur; this
refinement resulted in smaller deflections and much higher reaction forces. Parameters of the intramedullary rod studied were diameter and elastic modulus.

2.1.3. Objective

The objective of this study was to use BEF equations as simple models of an implanted femoral stem. The BEF equations were used to study the overall effects of changing parameters related to the implanted femoral stem system. The parameters studied were elastic modulus of the beam (femoral stem), elastic modulus of the foundation (femur or testing device), and length of the foundation (level of support provided by the femur or testing device). These results were used as a starting point for finite element studies of the same system.
2.2. Methods

Two BEF differential equations were used to model an implanted femoral stem [60]. The first differential equation was for a beam under axial compression:

\[ EI \left( \frac{d^4y}{dx^4} \right) + N \left( \frac{d^2y}{dx^2} \right) + ky = 0 \]

where

\[ E = \text{beam elastic modulus} \]
\[ I = \text{beam moment of inertia} \]
\[ N = \text{axial force} \]
\[ k = \text{foundation modulus} \]

This beam (Fig. 2.1.a) modelled the part of the femoral stem which was surrounded by bone or a testing device. This was called the shaft of the femoral stem. The second differential equation was for a cantilever beam under an axial and transverse force:

\[ \left( \frac{d^2y}{dx^2} \right) + (N/EI) y = - \left( \frac{P}{EI} \right) x \]

where

\[ E = \text{beam elastic modulus} \]
\[ I = \text{beam moment of inertia} \]
\[ N = \text{axial force} \]
\[ P = \text{transverse force} \]

This cantilever beam (Fig. 2.1.b) modelled the part of the femoral stem which was not surrounded by bone or a testing device. This was called the neck of the femoral stem, since at the highest foundation length studied, this beam modelled the femoral neck. At lower
foundation lengths, however, it modelled the femoral neck, plus the portion of the femoral stem not supported by bone, as would be seen if proximal resorption had occurred.

The loads, dimensions, and material properties for the BEF equations were chosen to model a femoral stem implanted into a femur or testing device. The compressive (N) and transverse (P) forces were 1200 N and 600 N respectively [51]. The moment (M₀) on the shaft was determined by the moment on the neck at the elastic foundation boundary. The beam radius was 8 mm. The total beam length was 190 mm, and the neck length was 50 mm. The foundation thickness was 8 mm. The elastic modulii investigated were 10, 20, 100, and 200 GPa; these modulii represented half the modulus of cortical bone, cortical bone, titanium, and cobalt-chrome.

Three variables were studied using the BEF models: beam elastic modulus, foundation elastic modulus, and foundation length. The elastic modulii studied were 10, 20, 100 and 200 GPa. For changes in beam elastic modulus, representing changes in femoral stem elastic modulus, the modulus of the foundation was fixed at 20 GPa. Likewise, for changes in foundation elastic modulus, representing changes in the modulus of the femur or testing device into which the femoral stem is implanted, the modulus of the beam was fixed at 20 GPa. In both cases, the foundation length was fixed at 140 mm; this was the highest foundation length studied. The foundation lengths studied were 50, 95, and 140 mm (low, medium and high embedding heights); the beam and foundation elastic modulii were fixed at 20
GPa (Fig. 2.2). The different foundation lengths modelled different
degrees of proximal resorption, as either present in the femur or
modelled in a testing device as different embedding heights.
\[ y = \left( \frac{M_0}{EI}D_2 \right) \left[ B (3a^2 - B^2) \sinh \alpha x \cos \beta x + \alpha (3B^2 - a^2) \sin \beta x \right] + \alpha \left[ B (\beta \sinh \beta x - \alpha \cosh \alpha x) \sin \beta x' - \alpha \cosh \alpha x \sin \beta x' \right] \]

\( D_1, D_2 = f(\alpha, \beta, L) \)
\( \alpha, \beta = f(k, N, EI) \)
\( x' = L - x \)

Fig. 2.1. Beam-on-elastic-foundation models for an implanted femoral stem. Loads, dimensions and material properties for the equations were chosen to model a femoral stem implanted into a femur or testing device.

a. Shaft of femoral stem - the part of the femoral stem surrounded by bone or a testing device
\[ y = A \sin cx - \left( \frac{P}{N} \right) x \]

\[ A = f(\alpha, \beta, c, L, EI, P, N) \]
\[ c = \left( \frac{N}{EI} \right)^{1/2} \]

Fig. 2.1. continued

b. Neck of femoral stem - the part of the femoral stem not surrounded by bone or a testing device
Fig. 2.2. Foundation lengths studied a in beam-on-elastic foundation model of an implanted femoral stem. Different foundation lengths modelled different degrees of proximal resorption in a femur, or different embedding heights in a testing device. The total stem length was 190 mm. The beam and foundation elastic modulii were 20 GPa.

Low embed - 50 mm
Mid embed - 95 mm
Hi embed - 140 mm
2.3. Results

2.3.1. Variation in Beam Modulus

As the beam elastic modulus decreased, displacement, shear, and moment on the shaft damped out over a smaller length (Fig. 2.3.a,c,d). The maximum normal shear of the beam increased (Fig. 2.3.d), as did the displacement (Fig. 2.3.a,b). The moment on the neck increased, but only slightly (Fig. 2.3.c). The moment on the shaft decreased, except for right at the elastic foundation edge (Fig. 2.3.c).

2.3.2. Variation in Foundation Modulus

As the elastic modulus of the foundation decreased, the displacement of the beam increased (Fig. 2.4.a,b). The moment on the neck hardly changed at all when the elastic modulus of the foundation was changed by an order of magnitude (Fig. 2.4.c). The moment on the shaft increased, except for right at the elastic foundation edge (Fig. 2.4.c), and the maximum normal shear decreased (Fig. 2.4.d). The distance over which displacement, normal shear, and moment damped out within the shaft increased (Fig. 2.4.a,c,d).

2.3.3. Variation in Foundation Length

As the foundation length decreased, the displacement, moment, and normal shear of the beam increased (Fig. 2.5). The distances over which the moment and shear damped out within the shaft were the same for the three foundation lengths (Fig. 2.5.b,c).
**Fig. 2.3.** Effects of varying stem elastic modulus in a beam-on-elastic-foundation model of an implanted femoral stem. The displacement, moment, and normal shear of the beam (stem) damped out over a smaller length as the stem elastic modulus decreased. Foundation elastic modulus was 20 GPa.

a. Displacement of stem, over proximal half of shaft. The displacement increased as the elastic modulus of the beam decreased.
Fig. 2.3. continued

b. Displacement of stem, at shaft/neck transition (embedding height). The displacement increased as the elastic modulus of the beam decreased.
Fig. 2.3. continued

c. Moment on stem, over neck and proximal half of shaft. The moment
on the neck increased only slightly as the elastic modulus of the
beam decreased. The moment on the proximal half of the shaft
decreased as the elastic modulus of the beam decreased, except for
right at the foundation edge.
Fig. 2.3. continued

d. Normal (interface) shear on stem, over proximal half of shaft. The maximum normal shear increased as the elastic modulus of the beam decreased.
Fig. 2.4. Effects of varying foundation elastic modulus in a beam-on-elastic-foundation model of an implanted femoral stem. The displacement, moment, and normal shear of the beam (stem) damped out over a longer length as the foundation elastic modulus decreased. Beam elastic modulus was 20 GPa.

a. Displacement of stem, over proximal half of shaft. The displacement increased as the elastic modulus of the foundation decreased.
Fig. 2.4. continued

b. Displacement of stem, at shaft/neck transition (embedding height). The displacement increased as the elastic modulus of the foundation decreased.
c. Moment on stem, over neck and proximal half of shaft. The moment on the neck hardly changed at all as the elastic modulus of the foundation decreased. The moment on the shaft increased, except for right at the foundation edge, as the elastic modulus of the foundation decreased.
d. Normal (interface) shear on stem, over proximal half of shaft. The maximum normal shear decreased as the elastic modulus of the foundation decreased.
Fig. 2.5. Effects of varying foundation length (embedding height) in a beam-on-elastic-foundation model of an implanted femoral stem. The displacement, moment, and normal shear of the beam (stem) increased as the foundation length decreased. Beam and foundation elastic modulii were 20 GPa.

a. Displacement of stem, at shaft/neck transition (embedding height).
b. Moment on stem, over entire shaft and neck. The distance over which the moment damped out within the shaft was the same for the three foundation lengths.
Fig. 2.5. continued

c. Normal (interface) shear on stem, over entire shaft. The distance over which the moment damped out within the shaft was the same for the three foundation lengths.
2.4. Discussion

2.4.1. Variation in Beam Modulus

A decrease in beam elastic modulus caused the displacement, shear, and moment on the shaft to damp out over a smaller length. In other words, the length of the femoral stem could be decreased if the elastic modulus of the stem is decreased. It has previously been observed that a femoral stem with a lower elastic modulus can be shorter than a stem with a higher elastic modulus [61].

A decrease in beam elastic modulus also caused the maximum normal shear to increase. The increase in shear with a decrease in femoral stem elastic modulus is a potential problem with polymer composite femoral stems [67].

As the beam elastic modulus decreased, the displacement of the beam increased. This indicated that a decrease in the elastic modulus of the femoral stem should cause an increase in the amount of displacement observed. The only slight increase in the moment on the neck could mean that the moment on a femoral stem neck has little dependence on the elastic modulus of the femoral stem. The moment on the shaft, however, decreased as beam elastic modulus decreased, which could mean that the moment on the shaft of the femoral stem decreases as the elastic modulus of the femoral stem decreases.

2.4.2. Variation in Foundation Modulus

BEF theory was used previously to study how an implanted femoral stem system changed with variations in the elastic foundation, where the foundation modelled the bone instead of a
testing device [64,65]. The BEF equations and models were different than the ones used here; one difference was that the earlier models included a cement layer. The range of bone modulii studied would also be smaller than the range of testing device modulii studied here. An increase in bone stiffness, either through increasing modulus or thickness, caused an overall decrease in cement, stem and interfacial stresses; proximal stresses increased, but only slightly [65]. Increasing the bone stiffness also extended the lengths over which load was transferred between the bone and the stem, both proximally and distally [64].

As the elastic modulus of the foundation decreased, the displacement of the beam increased. As interpreted through the case of a femoral stem, the displacement of the femoral stem increases as the elastic modulus of the testing device decreases.

For a high foundation length, the moment on the neck hardly changed at all when the elastic modulus of the foundation was changed by an order of magnitude. This could indicate that when a femoral stem is proximally potted in a testing device, the elastic modulus of the testing device has little or no effect on the stresses on the length of the femoral stem not embedded in the device.

The moment, maximum normal shear, and distance over which displacement, normal shear, and moment within the shaft damped out changed with a variation in foundation modulus. This indicated that the design of a cantilever-type femoral stem testing device affected the stresses seen on the femoral stem being tested.
2.4.3. Variation in Foundation Length

As the length of the foundation decreased, indicating less support to the beam, the displacement, moment, and normal shear of the beam increased. By analogy, the stresses in a femoral stem depend on the degree of proximal resorption, as either present in a bone or modelled in a testing device. A testing device with proximal potting (high embedding height) would result in higher stem stresses than would a testing device with distal potting (low embedding height).
2.5. Conclusions

Beam-on-elastic-foundation theory showed that the stresses on a simple femoral stem model depended on parameters of the testing device, as well as on parameters of the femoral stem. While BEF theory can indicate some effects of design changes to a testing device and femoral stem, the equations have some limitations. Distal stem behavior cannot always be modelled. Only simple loads and geometries can be used. Stress concentrations are not accounted for, and the equations represent perfectly bonded interfaces, which is not necessarily the case for implanted femoral stems. Results from these simpler models, however, can be used as a starting point for more complex finite element models [62]. The next step was to use finite element analysis to model an implanted polymer composite femoral stem.
3. Finite Element Analysis - Effects on Composite Stem

3.1. Introduction

3.1.1. Theory

Finite element analysis (FEA) involves breaking up a continuous system into a finite number of discrete elements [68,69]. Each element is described by a finite number of parameters. The solution of the continuous system is approximated from the assembly of the elements. Use of smaller and smaller elements, by increasing the number of elements used, can lead to better approximations of the continuous model.

One use for FEA is in stress analysis, when an exact analytical solution to a problem cannot be found because of complexities in material, geometry, loading, or boundary conditions [68,69]. Each finite element is described by a force-displacement relationship,

$$ [F] = [K] [u] $$

in which $F$ is the force vector, $u$ is the displacement vector, and $K$ is the element stiffness matrix. The forces and displacements are described only at nodes, which are discrete points within the elements. The element stiffness matrix is determined by the element material properties and the shape of the element used; it is analogous to the spring constant in the one-dimensional case. The element force-displacement relationships are assembled into a global force-displacement relationship, and then the nodal displacements are solved for. Displacements are usually interpolated for non-nodal
locations within the elements. The strains and the stresses for the model can then be calculated from the displacements.

Two techniques used in FEA to decrease the analysis time for three-dimensional (3-D) models are the two-dimensional (2-D) plus sideplates technique, and the 3-D axisymmetric technique. The 2-D plus sideplates technique involves making a 2-D mesh of a 3-D model, then accounting for the stiffness of the 3-D model by adding sideplates to the 2-D model [64]. Here, sideplates were added by increasing the thickness of the 2-D model elements (Fig. 3.1). The 3-D axisymmetric technique is used with models which are geometrically identical when rotated about an axis [69]. If the loading is axisymmetric, only one slice of the entire model is required for the finite element analysis. If the loading is nonaxisymmetric, more slices are necessary; the number of slices increases the accuracy of the analysis and can be chosen by the user in the finite element package ABAQUS [70]. The analysis time for axisymmetric models is considerably less than for the equivalent 3-D models. Both techniques have been used in FEA of total hip replacement [64,71-77].

3.1.2. Literature

In the field of total hip replacement, FEA has been used to compare different femoral stem geometries, materials, interfaces with the bone, and load angles, and has shown that these variables affect the stress state on the stem and the surrounding bone. Variables of stem geometry include length, cross-sectional area, and presence or
absence of a calcar collar [77-79]. Different types of interfaces between the stem and the femur were modelled differently; cemented or porous coating ingrowth femoral stems were usually modelled as fully bonded, while pressfit femoral stems were modelled with nonlinear gap or interface elements between the stem and the bone [71,74]. Varying the angle and magnitude of the applied load changed the loading situation on the stem [22]. In some studies, the elastic modulus of the stem was decreased to values which were both closer to the elastic modulus of the bone, and consistent with values for composite stems [67,76,78]. In other studies, not only was the elastic modulus of the stem decreased, but the anisotropy of the composite material was accounted for as well [35,40,49].

There are few finite element models of mechanical testing devices for femoral stems. Rohlmann et al. compared FEA and strain gage strains on the outer fixture surface of a proximal-potted testing device [22]. The strain gage strains were much higher than the FEA strains. This difference was attributed to interface loosening between the stem and cement in the testing device. Svensson et al. compared FEA and strain gage strains on a titanium stem in a distal-potted testing device [80]. With standard finite elements, the two sets of surface stresses were in close agreement in the nonembedded portion of the stem. When joint elements were added to the finite element model, allowing interfacial slip to occur, the two sets of surface stresses were in close agreement in both the embedded and nonembedded portions of the stem. Christel et al.
simulated distal potting in a finite element model by rigidly fixing
the distal portion of the stem [49]. The finite element model of a
carbon-carbon composite stem in this testing device confirmed the
direction and location of maximum stresses found in the femoral stem
in the actual distal-potted testing device. A mechanical testing
device for composite stems, developed by Humphrey and Gilbertson and
described in Section 1.1, was designed to reproduce the femoral head
deflections of a 3-D finite element beam model of the femoral stem
when well-fixed in the femur [19].

3.1.3. Objective

The objective of this study was to use finite element analysis
(FEA) to parametrically examine different configurations and
variables of a mechanical testing device for a polymer composite-
femoral stem. The changes in the composite stem and testing device
were compared to a baseline testing device. The changes in the
composite stem in the various testing devices were also compared to
the composite stem in a human femur, to find the best match.
Simplified finite element models were used to gain familiarity with
FEA and to help decide what parameters to study, before working with
full 3-D models.
\[ t = \text{area (A) / width = A for unit width} \]

\[ E_3 \cdot t_{\text{sideplate}} = \sum (E \cdot t) = E_1 t_1 + E_2 t_2 + E_2 t_3 + E_3 t_4 + E_2 t_5 + E_2 t_6 + E_1 t_7 \]

\[ t_{\text{sideplate}} = \frac{\sum (E \cdot t)}{E_3} \]

**Fig. 3.1.** Example of a sideplate calculation, for one element of unit width, in a three-dimensional cylindrical model.

**E = modulus of elasticity**  
**A = area**  
**t = thickness**
3.2. Methods

Three simplified and two full 3-D finite element models were constructed. These models, and parametric variations on these models, were analyzed. The simplified finite element models were a two-dimensional (2-D) plus sideplates stem-tester model, an axisymmetric stem-tester model, and an axisymmetric stem-bone model. The full 3-D finite element models were a stem-bone model and a stem-tester model. The 2-D plus sideplates model was analyzed using the finite element package ANSYS 4.4A (Swanson Analysis Systems, Inc.), on an IBM-PC; all other models were analyzed using the finite element package ABAQUS 5.2-1 (Hibbett, Karlsson and Sorensen, Inc.), on a Sun/SPARC workstation. All of the finite element models used quadratic elements.

3.2.1. Simplified Models

3.2.1.1. Two-Dimensional Plus Sideplates Stem-Tester Model

This model consisted of a cylindrical non-tapered femoral stem in a cylindrical testing device consisting of an outer fixture and an embedding compound, bone cement (Fig. 3.2). The stem, cement, and fixture were modeled as being perfectly bonded to one another. The materials were all isotropic, with a stem modulus of 25 GPa, a cement modulus of 3 GPa, and an initial fixture modulus of 100 GPa. The loading was an axial force of -1200 N and a transverse force of 600 N, applied to the top center of the stem [51].

Five variations in the testing device were studied using this model (Fig. 3.3). The variations from the baseline model were
fixture modulus reduced five times, fixture radius increased three times, distal fixture length increased, and embedding height decreased to distal potting.

3.2.1.2. Axisymmetric Stem-Tester Model

This model consisted of an axisymmetric model of a tapered femoral stem inserted in the same testing device as for the previous model (Fig. 3.4). The materials were modelled as being perfectly bonded to one another. The stem modulus was changed to 60 GPa. The cement modulus and initial fixture modulus were the same as for the two-dimensional plus sideplates model; 3 GPa and 100 GPa, respectively. The loading was a uniform axial pressure of 10 MPa applied to the top surface of the stem.

Eight variations in the testing device were studied (Fig. 3.5). The variations from the baseline model were fixture modulus reduced five times, fixture radius increased three times, distal fixture length increased, embedding height decreased to distal potting, rigid proximal potting, rigid distal potting, cement outer radius increased two times, and a tapered cement radius.

3.2.1.3. Axisymmetric Stem-Bone Model

This model consisted an axisymmetric model of a tapered femoral stem in a femur (Fig. 3.6). This model had interface elements between the stem and the bone, and inhomogeneous bone properties [52]. The stem/bone interface had a coefficient of friction of 0.4. The modulus of the cortical bone was 20 GPa. The modulus of the cancellous bone varied between 19.17 GPa distally and 5.833 GPa.
proximally. The loading was an axial pressure of 10 MPa and a transverse force of 600 N applied to the top surface of the stem.

Differences in interfacial shear stresses between stems of different materials were studied using this model. Proximal and distal interfacial shear stresses between the femoral stem and the bone were compared between isotropic stems of different modulii, and between an isotropic and transversely isotropic composite stem. The modulii of the isotropic stems were 200, 100, 60 and 20 GPa. These modulii were used to represent cobalt-chrome, titanium, polymer composite layup, and cortical bone. The isotropic stem having a modulus of 60 GPa was compared with a transversely isotropic stem having a longitudinal modulus of 60 GPa. The engineering constants for these two stems were as follows:

<table>
<thead>
<tr>
<th>Isotropic Stem</th>
<th>Transversely Isotropic Stem</th>
</tr>
</thead>
<tbody>
<tr>
<td>$E = 60 \text{ GPa}$</td>
<td>$E_L = 60 \text{ GPa}$</td>
</tr>
<tr>
<td>$v = 0.3 \text{ GPa}$</td>
<td>$E_T = 8.9 \text{ GPa}$</td>
</tr>
<tr>
<td>$G_{LT} = 5.1 \text{ GPa}$</td>
<td>$G_{LT} = 5.1 \text{ GPa}$</td>
</tr>
<tr>
<td>$v_{LT} = 0.3$</td>
<td>$v_{TT} = 0.045$</td>
</tr>
<tr>
<td>$L = \text{longitudinal direction (z)}$</td>
<td>$T = \text{transverse direction}$</td>
</tr>
</tbody>
</table>

An example of a transversely isotropic composite is a pultruded fiber bundle, which is isotropic in the $x$ and $y$ directions. This is in contrast to the more common orthotropic composite, such as a fabric or prepreg layup, which is anisotropic in all three structural directions. The transversely isotropic composite stem, however, could be modelled axisymmetrically, while an orthotropic composite stem could not.
3.2.2. Full 3-D Models

The full 3-D finite element models consisted of a composite stem implanted in either a human femur or a testing device. The material for the femoral stem was a quasi-isotropic carbon fiber/polysulfone layup, with the plies in the frontal plane (Hercules, Inc.). The material properties were proprietary. A quasi-isotropic layup has been used previously for composite stem FEA; its in-plane modulus was 51.8 GPa [35]. The finite element geometry of the composite stem was determined from a blueprint of the stem (Biomet, Inc.); the stem was symmetric and straight-stemmed (Fig. 3.7). The models had interface elements between the stem and the femur or testing device. These interfaces had an interference fit of 0.001 mm; this was a technique to get more reasonable interface behavior during FEA [81]. The applied load was distributed around the top edge of the stem. This load distribution modelled the "tip fit" seen when the head bore angle is greater than the stem cone angle. The models were analyzed using large displacement analysis. For the largest models, analysis time was about 3 hours.

The material properties and shape of the human femur were determined from a CT scan [52]. Simplifications for the finite element model of the human femur were that the geometry consisted of cylindrical and elliptical cross-sections, the bone properties were isotropic, and the cortical bone was homogeneous. The modulus of the cortical bone was 20 GPa. The modulus of the cancellous bone varied between 18.33 GPa distally and 5 GPa proximally. The coefficient of
friction between the stem and the testing device was 0.4; this value has been used previously in FEA of implanted composite stems [52].

The complete finite element model of the composite stem in the femur had 5155 nodes and 834 elements, including 124 interface elements (Fig. 3.8).

The baseline testing device was a cylindrical fixture with an outer radius of 45 mm and an inner radius of 35 mm. The modulus of the fixture was 70 GPa, to represent aluminum. The embedding compound between the fixture and the stem was bone cement, with a modulus of 3 GPa. The coefficient of friction between the stem and the bone cement was 0.4. The complete finite element model of the composite stem in the baseline testing device had 5705 nodes and 890 elements, including 132 interface elements (Fig. 3.9).

The finite element values generated were the strain components, averaged at the nodes. The value $(\varepsilon'_{xy} + \varepsilon'_{yz})^2$ was calculated as a measure of interlaminar shear strain (eil). The maximum values for the tensile strains in the x, y, and z directions, exx, eyy, and ezz, respectively, and eil were compared to values from the stem in the baseline tester, and to values from the stem in the femur. These bases of comparison were referred to as baseline values and femur values, respectively.

Parameters varied within the testing device model were applied load magnitude (l), fixture outer radius (rf), cement modulus (Ec), testing device modulus (Et), stem/cement interface friction (µ), applied load angle (a), and embedding height (z). Applied load
magnitude and angles were also varied within the femur model. The variations are listed below. The variations in boldface were those used in the baseline model; except when indicated otherwise, when one parameter was varied, all other parameters remained the same as for the baseline model.

1 - load magnitude (X BW)  2, 4, 6, 8
rf - fixture outer radius (mm)  45, 50, 55, 60
Ec - cement modulus (GPa)  0.5, 1, 3, 5, 10, 20
Et - testing device modulus (GPa)  3, 15.2*, 20, 70, 100, 200, \infty
\mu - interface friction  0.0, 0.2, 0.4, 0.8, 1.2, 1.6, 2.0
a - load angle designations  gait, 10/0, 10/9, 0/10
z - embedding height designations  z43, z39, z33, z29, z27, z25, z23

*orthotropic material

The baseline applied load was 4671 N. This was 6 times a body weight (BW) of 778 N (175 lbs.).

The baseline outer fixture radius of 45 mm had an inner radius of 35 mm; this resulted in a minimum bone cement thickness, or distance between the stem and the fixture, of 10 mm. This was the minimum embedding compound thickness recommended in ASTM F1440-92 [4]. The fixture thickness was kept constant at 10 mm.

The testing device moduli of 3, 15.2, 20, 70, 100 and 200 GPa represented testing devices made out of only delrin or acrylic, birchwood, isoelastic material, aluminum, titanium, and cobalt-chrome, respectively. There was no cement layer. The testing device modulus of \infty represented a completely rigid testing device. The birchwood was orthotropic instead of isotropic; its longitudinal modulus was 15.2 GPa, its radial modulus was 1.19 GPa, and its transverse modulus was 0.762 GPa [82]. Another hardwood, beech, has
been used as a femoral stem testing device, because of the wood's mechanical properties being close to those of cortical bone [25]. Two birchwood testing device finite element models were run - one model with the transverse and radial directions oriented in the x and y directions, respectively, and the other model with those orientations reversed. Since the resulting values for each orientation were fairly close, only the first orientation was used for comparison to the other testing devices.

The friction of the stem/cement or stem/bone interface is unknown [71,83]. The baseline value of $\mu = 0.4$ was a composite stem/bone interface friction used previously in FEA of implanted composite stems [52].

The baseline load angle represented a load angle seen during horizontal walking - 16.3° in-plane, and 11.8° out-of-plane [52]. The other load angles studied were three in-plane/out-of-plane angle combinations found in femoral stem testing standards and standard drafts - 10/0 [3-5], 10/9 [2,6], and 0/10 [29]. These load angles were designated at gait, 10/0, 10/9, and 0/10, respectively.

The embedding height designated as z43 represented full proximal support or proximal potting; the stem was supported up to the neck. The embedding height designated as z23 had 81.8 mm of the femoral stem unsupported, as measured from the top center of the stem. This embedding height was close to the unsupported femoral stem length of 80 mm, measured from the center of the head, specified in ISO standards involving distal potting of femoral stems [5,6].
The other embedding height designations were for intermediate embedding heights (Fig. 3.10), and represented different degrees of loosening or proximal resorption [7,8]. The numbers for the embedding height designations were nodal levels in the finite element model. For decreasing the embedding height, the fixture radii were also decreased to obtain a more constant degree of support, relative to the stem thickness at each height. The embedding heights were varied for all four load angles studied.
Fig. 3.2. Two-dimensional plus sideplates finite element model of a femoral stem implanted in a testing device.
Fig. 3.3. Variations in the testing device studied, using the two-dimensional plus sideplates model. Models are not to scale.
Fig. 3.4. Axisymmetric finite element model of a femoral stem implanted in a testing device.
Fig. 3.5. Variations in the testing device studied, using the axisymmetric model. Models are not to scale, and do not show the stem taper which was present in the finite element models.
Fig. 3.6. Axisymmetric finite element model of a femoral stem implanted in a femur.
Fig. 3.7. Three-dimensional finite element model of the composite stem.
Fig. 3.8. Three-dimensional finite element model of the composite stem in a human femur. Only half of the femur is shown.
Fig. 3.9. Three-dimensional finite element model of the composite stem in the baseline testing device. Only half of the testing device is shown.
Fig. 3.10. Embedding heights for the composite stem testing device. The embedding heights range between full proximal potting (z43) and distal potting close to that specified in ISO standards (z23) [5, 6].
3.3. Results

3.3.1. Simplified Models

3.3.1.1. Two-Dimensional Plus Sideplates Stem-Tester Model

The maximum von Mises stem stress was compared between stems in different testing devices (Table 3.1). Increasing the fixture radius decreased this stress. Increasing the length of the fixture below the femoral stem did not change this stress. Decreasing the modulus of the fixture increased this stress, and distal potting increased this stress the most.

For all variations in the testing device, the maximum von Mises stem stress was located where the femoral stem entered the testing device (Fig. 3.11). Changing the testing device from proximal to distal potting had an effect on the stress pattern, since the maximum von Mises stem stress was located at the lower embedding height. However, the variations of the testing device with proximal potting did not change the stress pattern much.

3.3.1.2. Axisymmetric Stem-Tester Model

With proximal potting, the variations in the testing device had very little effect on the maximum von Mises stem stress (Table 3.2). The stress decreased very slightly with an increased cement outer radius and a decreased fixture modulus. The stress increased very slightly with a tapered cement radius, an increased fixture outer radius, and rigid proximal potting. Rigid proximal potting increased this stress somewhat more than the other variations, but not
drastically. Distal potting of the femoral stem, either with a nonrigid or rigid testing device, increased this stress the most.

For the testing devices with proximal potting, the maximum von Mises stem stress was located at the junction of the straight part and the tapered part of the neck (Fig. 3.12.a). Increasing the device rigidity caused a change in the von Mises stress patterns of the stem (Fig. 3.13). Compared to the baseline model, decreasing the fixture modulus resulted in a distal shift in the transition to the lowest stress range. The completely rigid proximal potting resulted in a proximal shift in the transition to the lowest stress range. Increasing the fixture radius had less of an effect on the stress patterns; the transition to the lowest stress range shifted only slightly proximally.

For the testing devices with distal potting, the maximum von Mises stem stress was located where the femoral stem entered the tester (Fig. 3.12.b). There was also a stress concentration at the junction of the straight part and the tapered part of the neck.

3.3.1.3. Axisymmetric Stem-Bone Model

Changing the material of the femoral stem in this model affected the interfacial shear stress (IFSS) between the stem and the bone. As the elastic modulus of the stem was lowered, the proximal IFSS increased, and the distal IFSS decreased, as shown in Table 3.3:
<table>
<thead>
<tr>
<th>Stem Modulus (GPa)</th>
<th>IFSS (MPa) Proximal</th>
<th>IFSS (MPa) Distal</th>
</tr>
</thead>
<tbody>
<tr>
<td>200</td>
<td>2.11</td>
<td>2.46</td>
</tr>
<tr>
<td>100</td>
<td>2.47</td>
<td>1.90</td>
</tr>
<tr>
<td>50</td>
<td>2.90</td>
<td>1.43</td>
</tr>
<tr>
<td>20</td>
<td>4.53</td>
<td>0.60</td>
</tr>
</tbody>
</table>

Table 3.3. Variation in proximal and distal interfacial shear stress with change in stem modulus.

This was consistent with finite element results found in the literature, for fully bonded stem/bone interfaces [7,35,67,74].

Here, the stem/bone interface friction was \( \mu = 0.4 \). Changing from an isotropic stem (\( E = 60 \) GPa) to a transversely isotropic stem (\( E_{\text{longitudinal}} = 60 \) GPa), however, decreased both the proximal and distal IFSS's, as shown in Table 3.4:

<table>
<thead>
<tr>
<th>Stem Material</th>
<th>IFSS (MPa) Proximal</th>
<th>IFSS (MPa) Distal</th>
</tr>
</thead>
<tbody>
<tr>
<td>Isotropic</td>
<td>2.90</td>
<td>1.43</td>
</tr>
<tr>
<td>Trans. iso.</td>
<td>2.72</td>
<td>1.03</td>
</tr>
</tbody>
</table>

Table 3.4. Variation in proximal and distal interfacial shear stress with change in stem anisotropy.
Fig. 3.11. Contour plot of von Mises stresses for the two-dimensional plus sideplates stem-tester baseline model. Only the upper part of the model is shown. Discontinuous contours were necessary because of the sideplates. For all the testing device models studied, the maximum von Mises stress was located where the stem entered the tester.
Fig. 3.12. Contour plots of von Mises stresses for axisymmetric stem-tester models with proximal and distal potting. For the tester models with proximal potting, the maximum von Mises stress was located at the junction of the straight part and the tapered part of the neck. For the tester models with distal potting, the maximum von Mises stress was located where the femoral stem entered the tester.

Scale (in MPa): Red 10.2 -> Orange 8.17 -> 10.2 Yellow 6.14 -> 8.17 Lt. green 4.10 -> 6.14 Lt. blue 2.07 -> 4.10 Blue 0.00 -> 2.07
Fig. 3.13. Contour plot of von Mises stem stresses for axisymmetric femoral stems in testing devices of increasing rigidity. Increasing the testing device rigidity caused a proximal shift in the transition to the lowest stress range in the stem.

axi3 - fixture modulus reduced 5X
axi4 - baseline model
axi5 - fixture radius increased 3X
axi8 - rigid proximal potting

Scale (in MPa):

<table>
<thead>
<tr>
<th>Color</th>
<th>Scale 1</th>
<th>Scale 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Red</td>
<td>9.54</td>
<td></td>
</tr>
<tr>
<td>Orange</td>
<td>7.63</td>
<td>9.54</td>
</tr>
<tr>
<td>Yellow</td>
<td>5.72</td>
<td>7.63</td>
</tr>
<tr>
<td>Lt. green</td>
<td>3.81</td>
<td>5.72</td>
</tr>
<tr>
<td>Lt. blue</td>
<td>1.90</td>
<td>3.81</td>
</tr>
<tr>
<td>Blue</td>
<td>0.00</td>
<td>1.90</td>
</tr>
</tbody>
</table>
Table 3.1. Maximum von Mises femoral stem stress with the stem in different testing devices, in a two-dimensional plus sideplates finite element model.

<table>
<thead>
<tr>
<th>Testing Device Variation</th>
<th>Max. von Mises Stress (MPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fixture outer radius increased 3X</td>
<td>14.241</td>
</tr>
<tr>
<td>Baseline model</td>
<td>17.793</td>
</tr>
<tr>
<td>Distal fixture length increased</td>
<td>17.793</td>
</tr>
<tr>
<td>Fixture modulus reduced 5X</td>
<td>22.237</td>
</tr>
<tr>
<td>Distal potting (50 mm)</td>
<td>59.358</td>
</tr>
</tbody>
</table>

Table 3.2. Maximum von Mises femoral stem stress with the stem in different testing devices, in an axisymmetric finite element model.

<table>
<thead>
<tr>
<th>Testing Device Variation</th>
<th>Max. von Mises Stress (MPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cement outer radius increased 2X</td>
<td>268.94</td>
</tr>
<tr>
<td>Fixture modulus reduced 5X</td>
<td>269.10</td>
</tr>
<tr>
<td>Baseline model</td>
<td>269.28</td>
</tr>
<tr>
<td>Tapered cement radius</td>
<td>269.31</td>
</tr>
<tr>
<td>Fixture outer radius increased 3X</td>
<td>269.34</td>
</tr>
<tr>
<td>Rigid proximal potting</td>
<td>274.06</td>
</tr>
<tr>
<td>Distal potting (50 mm)</td>
<td>281.41</td>
</tr>
<tr>
<td>Rigid distal potting (50 mm)</td>
<td>451.98</td>
</tr>
</tbody>
</table>
3.3.2. Full 3-D Models

The finite element values used for analyzing effects of the testing device on the femoral stem were the maximum tensile stem strains in the global x, y and z directions (exx, eyy and ezz) and interlaminar shear strain (eil), (Table 3.5). The x direction was medial-lateral, the y direction was anterior-posterior, and the z direction was proximal-distal. These strains were normalized to the stem strains from the baseline tester (baseline values), and to the stem strains from the femur (femur values).

The strains studied would lead to different failure modes within the composite. The composite plies were in the x-z plane, so the stem strains exx and ezz would lead to tensile failure within the composite planes. Stem strains eyy and eil would lead to interlaminar failure, by normal tensile strain and shear strain, respectively.

3.3.2.1. Comparisons to Baseline Tester Values

The maximum stem strains exx, eyy, ezz, and eil were normalized to the baseline tester values. The maximum values were used, regardless of their node locations. For the testing device variations with proximal potting, the node location for maximum stem eil occasionally changed. These changes are indicated on the figures which follow. When the embedding height was lowered, the node location for the maximum stem strains often switched from the neck to around the lower embedding height. The crossover was defined as the two embedding heights between which this switch occurred. Maximum
stem strains which were below the crossover are indicated on the figures which follow.

The maximum stem strains were linear ($r^2 > 0.998$) over the load range studied. This was in spite of the nonlinearity of the full finite element model.

As the outer fixture radius increased to 60 mm, only maximum stem eil changed noticeably, increasing to 109% of the baseline value (Fig. 3.14). The maximum stem eil was below the neck.

As the cement modulus ($E_c$) increased, only maximum stem ezz and eil showed much change; at the highest $E_c$, maximum stem eil increased to 124% of the baseline value, and maximum stem ezz increased to 108% of the baseline value (Fig. 3.15). The node location for maximum stem eil was different when $E_c = 0.5$ and 1 GPa.

As the stem/cement friction ($\mu$) increased, only maximum stem eyy and eil showed much change; at $\mu = 2.0$, they increased to 106% and 105% of the baseline values, respectively (Fig. 3.16). Except for eil, the maximum stem strains started to level off at $\mu > 0.8$. The node location for maximum stem eil changed when $\mu \geq 1.2$.

The load angle 0/10 resulted in the highest maximum stem strains for most of the embedding heights and strains (Fig. 3.17). The load angle 10/0 resulted in the lowest maximum stem strains for most of the embedding heights and strains.

The location of the maximum stem exx stayed within the neck, for all load angles; there was no crossover (Fig. 3.17a). The crossover for maximum stem eyy was between the third and fourth
highest embedding heights, for all load angles (Fig. 3.17.b). The
crossover for maximum stem exz was between the second and third
highest embedding heights, for all load angles (Fig. 3.17.c). The
crossover for maximum stem eil was between the third and fourth
lowest embedding heights for load angles 10/9 and gait, between the
second and third lowest embedding heights for load angle 0/10, and
between the lowest and second lowest embedding heights for load angle
10/0 (Fig. 3.17.d).

Maximum stem exx increased steadily with decreased embedding
height (Fig. 3.17.a). There was hardly any change in maximum stem
eyy with embedding height, until after the crossover (Fig. 3.17.b).
There was not much change in maximum stem exz with embedding height
until after the crossover; however, the crossover was at a high
embedding height (Fig. 3.17.c). Maximum stem eil increased with
decreased embedding height, both before and after the crossover,
except for load angle 10/0 (Fig. 3.17.d). The increase in maximum
stem eil was steeper after the crossover.

The maximum stem strains in the neck were also recorded, to see
how these strains changed with embedding height (Fig. 3.18). The
node locations were taken as those for the maximum strains at the
highest embedding height. Maximum stem eil showed the most change;
it increased by nearly 2.5 times between the highest and lowest
embedding height. Maximum stem exx showed the least change; it
increased 14% between the highest and lowest embedding height.
Maximum stem eyy decreased at the lowest embedding height, after staying fairly constant for the other embedding heights.

The maximum stem strains for the highest and second highest embedding heights were compared (Table 3.6). The stem strains at the second highest embedding height showed the effects of a small decrease from full proximal potting. Maximum stem exx increased very slightly, maximum stem eyy decreased and had a node location change, maximum stem ezz decreased, and maximum stem eil increased and had a node location change.

3.3.2.2. Comparisons to Femur Values

The maximum stem strains exx, eyy, ezz, and eil were normalized to the femur values. The node locations for maximum stem exx and ezz were the same for the stem in the femur and for the stem in all testing device variations at the highest embedding height and load angle gait. However, the node locations for maximum stem eyy and eil were different for the stem in the femur and the stem in the testing device variations. For normalization to the femur values, the tester values at the node location for the maximum femur values were used.

The ranking of the tester values at the node location for the maximum femur values, relative to the maximum stem strains in the tester, were also recorded (Table 3.5). Any ranking other than one indicated a mismatch between node locations of maximum stem strains in the tester and maximum stem strains in the femur. A larger ranking indicated a larger mismatch between these node locations.
As the outer fixture radius increased, only maximum stem exx was close to the femur value; the other values were much lower (Fig. 3.19). Only maximum stem eyy changed much with fixture radius. The ranking of the tester eyy increased from 5 to 13. The ranking of the tester eil was 2.

As the cement modulus (Ec) increased, maximum stem exx stayed close to the femur value, maximum stem eyy started lower and decreased considerably, and maximum stem ezz and eil leveled off below the femur values (Fig. 3.20). The ranking of the tester eyy increased from 2 to 33. The ranking of the tester eil increased from 1 to 3. The node location for maximum eil was the same for the femur and the testing devices with Ec = 0.5 and 1 GPa.

As the testing device modulus (Et) increased, the maximum stem strains leveled off (Fig. 3.21). Maximum stem exx and ezz approached the femur values, and were close to the femur values for most of the testing device modulii studied. Maximum stem eyy decreased from 74% to 38% of the femur value, with the exception of the orthotropic birchwood testing device, which increased to 82% of the femur value. Maximum stem eil leveled off to 72% of the femur value. The maximum values for Et = 3 GPa and the orthotropic birchwood testing device (longitudinal modulus = 15.2 GPa) were very close for exx, ezz and eil. The maximum stem strains in a completely rigid testing device (Et = ∞), not included in the graph, followed the trends for increasing testing device modulus (Table 3.5). The ranking of the tester eyy increased from 3 to 113, except for the birchwood testing
device, which had a ranking of 2. The ranking of the tester eil was 2 for Et = 3, 15.2, and 20 GPa, and 4 for Et = 70, 100, 200, and ∞ GPa.

As the stem/cement interface friction (μ) increased, maximum stem exx was close to the femur value, maximum stem eyy started lower and decreased, and maximum stem ezz and eil leveled off at values below the femur values (Fig. 3.22). The ranking of the tester eyy increased from 2 to 14. The ranking of the tester eil was 3 at μ = 0.8 and 1.2, and 2 otherwise.

The maximum strains for the stem in the baseline tester were divided by the maximum strains for the stem in the femur, for each load angle. The percentage difference of these ratios were added, for the stem strains at the maximum node locations for the baseline testing device (Table 3.7.a), and at the maximum node locations for the femur (Table 3.7.b). This was to determine which load angle resulted in the best match between the composite stem in the tester and the composite stem in the femur. The two sets of maximum node locations frequently did not match. For the stem strains at the maximum node location for the tester, the best match was for 10/0, followed closely by 10/9 and gait. Load angle 0/10 resulted in the worst match. For the stem strains at the maximum node location for the femur, the best match was for 10/0, followed by 0/10, then gait. Load angle 10/9 resulted in the worst match, but ezz for this load angle had a sign change which drove up the percentage difference.
Three measures of the best choice of testing device, with regards to the maximum stem strains, were taken, since all of the testing devices fell short with regards to simulating the femur. The measures were taken from the strains normalized to the femur values; these normalized strains are seen in Figures 19-22. The first measure was the sum of the percentage differences of the strains normalized to the femur values. The smallest numbers indicated the smallest differences between the stem strains in the tester and the stem strains in the femur. The second and third measures were the stem strains eyy and eil normalized to the femur values. These last two measures were chosen because these stem strains were lower than the femur values, when compared to the stem strains exx and ezz. The best testing device variations, as per these three measures, were as follows (Table 3.8):

<table>
<thead>
<tr>
<th></th>
<th>Normalized eyy</th>
<th>Normalized eil</th>
</tr>
</thead>
<tbody>
<tr>
<td>Best</td>
<td>Birch</td>
<td>Ec=0.5</td>
</tr>
<tr>
<td>2nd best</td>
<td>μ=0.0</td>
<td>Birch</td>
</tr>
<tr>
<td>3rd best</td>
<td>Ec=0.5</td>
<td>Et=3</td>
</tr>
<tr>
<td>4th best</td>
<td>Ec=1</td>
<td>Ec=1</td>
</tr>
</tbody>
</table>

Table 3.8. Best choice of composite stem testing device. The choices listed were the variations from the baseline tester described in Section 3.2.2. Birch = birchwood testing device, μ = stem/cement interface friction, Ec = cement modulus (GPa) and Et = testing device modulus (GPa).

Based on these measures, birchwood was chosen as the best testing device. Beechwood, which is similar to birchwood, has been used previously as a femoral stem testing device [25].
The contour plots of stem strains in the birchwood testing device differed to some extent from the contour plots of stem strains in the femur. The contour plots of stem exx in the femur, baseline testing device, and birchwood testing device were very similar (Fig. 3.23). None of the testing device variations had much effect on maximum stem exx, but the birchwood testing device slightly improved this value over the baseline testing device. The contour plots of stem eyy in the femur, baseline testing device, and birchwood testing device were also similar, more so in the neck than below the neck, but the maximum stem eyy in the femur had a different node location (Fig. 3.24). The birchwood testing device improved the maximum stem eyy, and also improved the ranking: the maximum stem eyy for the birchwood testing device was at the node location for the second highest stem eyy for the femur. The contour plots of stem ezz in the baseline testing device and birchwood testing device were more similar to each other than to the contour plot of stem ezz in the femur, in the neck (Fig. 3.25). The contour plots of stem ezz in the femur, baseline testing device, and birchwood testing device were all different below the neck. The birchwood testing device slightly improved maximum stem ezz over the baseline testing device. No contour plot of stem e1l was available.

The parametric studies indicated that maximum stem eyy could be increased at the node locations for the maximum femur values by decreasing the fixture radius, the stem/cement interface friction, and possibly the longitudinal modulus of an orthotropic testing
device. The latter two parameters were also predicted to increase maximum stem eil. These parameter changes were predicted to cause either no change or slight improvement in the other maximum strains. The parameters fixture radius, stem/tester interface friction, and longitudinal modulus were changed in the finite element model of the birchwood testing device. The outer radius of the birchwood testing device was decreased from 45 mm to 40 mm and 35 mm. The element geometry was changed slightly from the original model for an outer fixture radius of 45 mm, to accommodate the decreased tester radii. The changes in the maximum stem strains were negligible between the old and new element geometries at 45 mm; however, for the new element geometry, the node location for maximum stem eil in the birchwood testing device now matched the node location for maximum stem eil in the femur. The stem/tester interface friction was decreased from \( \mu = 0.4 \) to 0.2 and 0.0 for the outer tester radius of 35 mm. The testing device material was changed to yellow poplar, which has a lower longitudinal modulus than birchwood [82].

These changes to the birchwood testing device gave the expected results: maximum stem eyy and eil slightly improved with respect to the femur values, and maximum stem exx and ezz did not change much (Table 3.9). The maximum node location for the stem in the testing device frequently matched the maximum node location for the stem in the femur, as indicated by the rankings (Table 3.10). The rankings were 1 for all maximum stem exx and ezz. The rankings were 1 for maximum stem eyy in the yellow poplar testing device and in the
birchwood testing device at $\mu = 0.2$ and 0.0 (outer tester radius = 35 mm). The rankings were 1 for maximum stem eil in the birchwood testing device at all outer fixture radii ($\mu = 0.4$), the birchwood testing device at $\mu = 0.2$ (outer tester radius = 35 mm), and the yellow poplar testing devices.
Max. Stem Strains, Vary Fixture Radius

Fig. 3.14. Effects of fixture outer radius on maximum stem strains in composite stem testing device finite element model. Fixture inner radius was 10 mm less than the fixture outer radius. Strains were normalized to the baseline testing device values.
Fig. 3.15. Effects of cement modulus on maximum stem strains in composite stem testing device finite element model. Strains were normalized to the baseline testing device values. The darker symbols indicated a change in node location.
Fig. 3.16. Effects of stem/cement interface friction on the maximum stem strains in composite stem testing device finite element model. Strains were normalized to the baseline testing device values. The darker symbols indicated a change in node location for maximum eil.
Fig. 3.17. Effects of embedding height and load angle on the maximum stem strains in composite stem testing device finite element model. Strains were normalized to the baseline testing device values for load angle gait. The darker symbols indicated maximum strains which occurred around the embedding height, rather than in the neck.

a. Maximum stem $\varepsilon_{xx}$
b. Maximum stem $\varepsilon_{yy}$
c. Maximum stem $\varepsilon_{zz}$
d. Maximum stem $\varepsilon_{il}$
Maximum Stem eyy, Vary Embedding Height

Fig. 3.17. continued
Maximum Stem ezz, Vary Embedding Height

![Graph showing Maximum Stem ezz, Vary Embedding Height with different symbols representing different values.](image)

**Fig. 3.17.** continued
Maximum Stem eil. Vary Embedding Height

![Graph showing normalized interlaminar shear strain vs. embedding height at stem midline (mm)]

Fig. 3.17. continued
Fig. 3.18. Effects of embedding height on the maximum stem strains in the neck in composite stem testing device finite element model. Strains were normalized to the baseline testing device values.
Fig. 3.19. Effects of fixture outer radius on the maximum stem strains in composite stem testing device finite element model. Fixture inner radius was 10 mm less than the fixture outer radius. Strains were normalized to the femur values.
Fig. 3.20. Effects of cement modulus on maximum stem strains in composite stem testing device finite element model. Strains were normalized to the femur values.
Max. Stem Strains, Vary Tester Modulus

Fig. 3.21. Effects of testing device modulus on the maximum stem strains in composite stem testing device finite element model. The testing device material at 15.2 GPa was birchwood, an orthotropic material; 15.2 GPa was the longitudinal modulus. The transverse modulus was 0.762 GPa and the radial modulus was 1.19 GPa. Strains were normalized to the femur values.

○ = exx birch
× = eyy birch
● = ezz birch
▲ = eil birch
Fig. 3.22. Effects of stem/cement interface friction on the maximum stem strains in composite stem testing device finite element model. Strains were normalized to the femur values.
Fig. 3.23. Contour plot of composite stem strains in the x-direction, with the stem in a femur, baseline testing device, and birchwood testing device, which was chosen as the best testing device. The scale was the same for all plots. The maximum stem exx in the femur, baseline testing device, and birchwood testing device occurred at the same location.

<table>
<thead>
<tr>
<th>Scale (in 1E-3):</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Red</td>
<td>4.46</td>
<td>-&gt;</td>
</tr>
<tr>
<td>Orange</td>
<td>3.55</td>
<td>4.46</td>
</tr>
<tr>
<td>Yellow-orange</td>
<td>2.65</td>
<td>3.55</td>
</tr>
<tr>
<td>Yellow</td>
<td>1.75</td>
<td>2.65</td>
</tr>
<tr>
<td>Green</td>
<td>0.85</td>
<td>1.75</td>
</tr>
<tr>
<td>Lt. green</td>
<td>-0.06</td>
<td>0.85</td>
</tr>
<tr>
<td>Lt. blue</td>
<td>-0.96</td>
<td>-0.06</td>
</tr>
<tr>
<td>Blue</td>
<td>-1.86</td>
<td>-0.96</td>
</tr>
<tr>
<td>Dk. blue</td>
<td></td>
<td>-1.86</td>
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</table>
Fig. 3.24. Contour plot of composite stem strains in the y-direction, with the stem in a femur, baseline testing device, and birchwood testing device, which was chosen as the best testing device. The scale was the same for all plots. The maximum stem eye in the femur occurred at the red spot to the right, which was at the resection level. The maximum stem eye in the testing devices occurred at the only visible red spot, which was above the potting level.

Scale (in 1E-3):  

<table>
<thead>
<tr>
<th>Color</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Red</td>
<td>1.57</td>
</tr>
<tr>
<td>Orange</td>
<td>1.01</td>
</tr>
<tr>
<td>Yellow-orange</td>
<td>0.44</td>
</tr>
<tr>
<td>Yellow</td>
<td>-0.12</td>
</tr>
<tr>
<td>Green</td>
<td>-0.68</td>
</tr>
<tr>
<td>Lt. green</td>
<td>-1.25</td>
</tr>
<tr>
<td>Lt. blue</td>
<td>-1.81</td>
</tr>
<tr>
<td>Blue</td>
<td>-2.38</td>
</tr>
<tr>
<td>Dk. blue</td>
<td>-2.38</td>
</tr>
</tbody>
</table>
Fig. 3.25. Contour plot of composite stem strains in the z-direction, with the stem in a femur, baseline testing device, and birchwood testing device, which was chosen as the best testing device. The scale was the same for all plots. The maximum stem ez in the femur, baseline testing device, and birchwood testing device occurred at the same location.

Scale (in 1E-3):

- Red: 1.44
- Orange: 0.65
- Yellow-orange: -0.15
- Yellow: -0.95
- Green: -1.75
- Lt. green: -2.54
- Lt. blue: -3.34
- Blue: -4.14
- Dk. blue: -4.14
Table 3.5. Maximum stem strains for a composite stem in a human femur and in testing devices with proximal potting. The testing device variations were in outer fixture radius (rf), cement modulus (Ec), testing device modulus (Et), and stem/cement interface friction (u). The tensile strains in the x, y and z directions are designated as exx, eyy and ezz. The interlaminar shear strain is designated as eil.

The node locations were usually different between the maximum stem strains eyy and eil in the testing devices and in the femur. The stem strains in the testing devices at the node locations for the maximum stem strain in the femur were recorded in the columns marked "per max femur." The ranking of how far these stem strains were below the maximum stem strains in the testing device were recorded in the columns marked "rank." Any ranking other than one indicated a mismatch between node locations of maximum stem strains in the tester and maximum stem strains in the femur.

<table>
<thead>
<tr>
<th>Strains (ms)</th>
<th>exx</th>
<th>eyy</th>
<th>eil</th>
</tr>
</thead>
<tbody>
<tr>
<td>femur</td>
<td>5.3615</td>
<td>2.1355</td>
<td>2.2417</td>
</tr>
<tr>
<td>per max femur</td>
<td>5.3615</td>
<td>2.1355</td>
<td>2.2417</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>rf (mm)</th>
<th>exx</th>
<th>eyy</th>
<th>eil</th>
<th>eyy rank</th>
<th>eil rank</th>
</tr>
</thead>
<tbody>
<tr>
<td>45</td>
<td>1.6266</td>
<td>1.7615</td>
<td>9.8136</td>
<td>1.4872</td>
<td>5  7.6228</td>
</tr>
<tr>
<td>55</td>
<td>1.3732</td>
<td>1.9587</td>
<td>3.2711</td>
<td>1.4411</td>
<td>7  7.6482</td>
</tr>
<tr>
<td>65</td>
<td>1.2632</td>
<td>1.7957</td>
<td>9.5112</td>
<td>1.2767</td>
<td>10 7.6536</td>
</tr>
<tr>
<td>75</td>
<td>1.2644</td>
<td>1.6457</td>
<td>3.7544</td>
<td>1.3463</td>
<td>12 7.6484</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Ec (GPa)</th>
<th>exx</th>
<th>eyy</th>
<th>eil</th>
<th>eyy rank</th>
<th>eil rank</th>
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</thead>
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<td>8.5</td>
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<td>0.8477</td>
<td>1.7152</td>
</tr>
<tr>
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<td>1.0165</td>
<td>1.9805</td>
<td>7.6482</td>
<td>1.6543</td>
</tr>
<tr>
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<td>1.0266</td>
<td>1.9616</td>
<td>0.8196</td>
<td>1.4822</td>
</tr>
<tr>
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<td>1.0221</td>
<td>2.0022</td>
<td>0.6576</td>
<td>1.3864</td>
</tr>
<tr>
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<td>1.0322</td>
<td>2.4622</td>
<td>9.3590</td>
<td>1.1767</td>
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<td>5.2165</td>
<td>1.0565</td>
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</tr>
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<td>$E_t$ (GPa)</td>
<td>$E_{xx}$</td>
<td>$E_{yy}$</td>
<td>$E_{zz}$</td>
<td>$G_{xx}$</td>
<td>$G_{yy}$</td>
</tr>
<tr>
<td>------------</td>
<td>---------</td>
<td>---------</td>
<td>---------</td>
<td>---------</td>
<td>---------</td>
</tr>
<tr>
<td>7</td>
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<td>1.973</td>
<td>0.400</td>
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<td>2.179</td>
<td>0.499</td>
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<tr>
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<tr>
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<td>2.217</td>
<td>0.577</td>
<td>1.262</td>
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<tr>
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<td>2.188</td>
<td>2.463</td>
<td>0.629</td>
<td>1.241</td>
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<table>
<thead>
<tr>
<th>$u$</th>
<th>$E_{xx}$</th>
<th>$E_{yy}$</th>
<th>$E_{zz}$</th>
<th>$G_{xx}$</th>
<th>$G_{yy}$</th>
<th>$G_{zz}$</th>
<th>$v_{xx}$</th>
<th>$v_{yy}$</th>
<th>$v_{zz}$</th>
<th>$v_{xy}$</th>
<th>$v_{xz}$</th>
<th>$v_{yz}$</th>
<th>$v_{el}$</th>
<th>$v_{rel}$</th>
<th>$v_{rel}$</th>
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<td>0.898</td>
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<td>0.333</td>
<td>1.736</td>
<td>0.333</td>
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<td>1.744</td>
<td>1.829</td>
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<td>0.333</td>
<td>0.929</td>
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<td>0.333</td>
<td>1.744</td>
<td>0.333</td>
<td>1.744</td>
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<td>1.808</td>
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<td>0.333</td>
<td>0.946</td>
<td>1.760</td>
<td>1.760</td>
<td>0.333</td>
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<td>1.760</td>
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<td>0.980</td>
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<td>1.787</td>
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<td>0.333</td>
<td>1.018</td>
<td>1.826</td>
<td>1.826</td>
<td>0.333</td>
<td>1.826</td>
<td>0.333</td>
<td>1.826</td>
</tr>
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<td>0.333</td>
<td>1.054</td>
<td>1.854</td>
<td>1.854</td>
<td>0.333</td>
<td>1.854</td>
<td>0.333</td>
<td>1.854</td>
</tr>
<tr>
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<td>5.152</td>
<td>2.383</td>
<td>1.902</td>
<td>1.094</td>
<td>1.884</td>
<td>1.787</td>
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<td>0.333</td>
<td>1.094</td>
<td>1.884</td>
<td>1.884</td>
<td>0.333</td>
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<td>1.884</td>
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<td>1.970</td>
<td>1.151</td>
<td>1.929</td>
<td>1.787</td>
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<td>0.333</td>
<td>1.151</td>
<td>1.929</td>
<td>1.929</td>
<td>0.333</td>
<td>1.929</td>
<td>0.333</td>
<td>1.929</td>
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<td>5.104</td>
<td>2.617</td>
<td>2.012</td>
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<td>1.946</td>
<td>1.787</td>
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<td>0.333</td>
<td>1.179</td>
<td>1.946</td>
<td>1.946</td>
<td>0.333</td>
<td>1.946</td>
<td>0.333</td>
<td>1.946</td>
</tr>
</tbody>
</table>
Table 3.6. Effect of slightly decreasing the embedding height of a composite stem in a testing device, from full proximal potting. The tensile strains in the x, y and z directions are designated as exx, eyy and ezz. The interlaminar shear strain is designated as eil. Numbers with asterisks indicate that the maximum stem strain occurred at a different node location between the two embedding heights.

<table>
<thead>
<tr>
<th>Embedding Height</th>
<th>exx</th>
<th>eyy</th>
<th>ezz</th>
<th>eil</th>
</tr>
</thead>
<tbody>
<tr>
<td>highest</td>
<td>5.2725</td>
<td>1.8266</td>
<td>1.9616</td>
<td>8.0186*</td>
</tr>
<tr>
<td>2nd highest</td>
<td>5.3371</td>
<td>1.7591*</td>
<td>1.7884</td>
<td>8.5073*</td>
</tr>
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</table>
Table 3.7. Comparison of maximum stem strains in a femur and the baseline testing device, at the maximum node locations for a) the baseline testing device and b) the femur. The designation gait indicates a load angle seen during horizontal walking. The designations 10/0, 10/9 and 0/10 indicate the in-plane/out-of-plane load angles. The tensile strains in the x, y and z directions are designated as exx, eyy and ezz. The interlaminar shear strain is designated as eil. The column marked "match" was the sum of the percentage difference of the strain ratios, used to determine which load angle resulted in the best match between the composite stem in the tester and the composite stem in the femur.

<table>
<thead>
<tr>
<th>Gait</th>
<th>Strains (1E-3)</th>
<th>exx</th>
<th>eyy</th>
<th>ezz</th>
<th>eil</th>
<th>match</th>
</tr>
</thead>
<tbody>
<tr>
<td>Femur</td>
<td></td>
<td>5.3613</td>
<td>1.9363</td>
<td>2.2417</td>
<td>6.6717</td>
<td></td>
</tr>
<tr>
<td>24°</td>
<td></td>
<td>5.2702</td>
<td>1.8268</td>
<td>1.9616</td>
<td>8.0136</td>
<td></td>
</tr>
<tr>
<td>24°/3°</td>
<td></td>
<td>0.963</td>
<td>0.963</td>
<td>0.863</td>
<td>1.263</td>
<td>0.33</td>
</tr>
</tbody>
</table>

| Femur  |                | 5.6162 | 1.8762 | 1.7862 | 3.2376 |       |
| 24°    |                | 5.3762 | 1.8268 | 1.7768 | 3.9776 |       |
| 24°/3° |                | 0.963 | 0.963 | 0.863 | 1.263 | 0.34 |

| Femur  |                | 6.2055 | 1.8055 | 1.6755 | 5.6755 |       |
| 24°    |                | 6.0755 | 1.8255 | 1.6255 | 8.3755 |       |
| 24°/3° |                | 0.963 | 0.963 | 0.863 | 1.263 | 0.35 |

| Femur  |                | 5.3854 | 2.0054 | 2.0354 | 11.3854 |       |
| 24°    |                | 5.7554 | 2.1854 | 2.0554 | 8.0154 |       |
| 24°/3° |                | 0.941 | 1.941 | 1.141 | 0.781 | 0.4 |
### Strains (1E-3)

<table>
<thead>
<tr>
<th>gait</th>
<th>exx</th>
<th>eyy</th>
<th>ezz</th>
<th>eil</th>
<th>match</th>
</tr>
</thead>
<tbody>
<tr>
<td>femur</td>
<td>5.3615</td>
<td>2.1365</td>
<td>3.2417</td>
<td>10.2940</td>
<td></td>
</tr>
<tr>
<td>z43</td>
<td>5.2726</td>
<td>1.4830</td>
<td>1.9616</td>
<td>7.6326</td>
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</tr>
<tr>
<td>z43/fem</td>
<td>0.98</td>
<td>0.69</td>
<td>0.65</td>
<td>0.74</td>
<td>0.71</td>
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</table>

### 10° W

<table>
<thead>
<tr>
<th>gait</th>
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<th>ezz</th>
<th>eil</th>
<th>match</th>
</tr>
</thead>
<tbody>
<tr>
<td>femur</td>
<td>5.3128</td>
<td>1.3770</td>
<td>1.7307</td>
<td>3.7266</td>
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</tr>
<tr>
<td>z43</td>
<td>5.2983</td>
<td>1.7567</td>
<td>1.7727</td>
<td>3.6220</td>
<td></td>
</tr>
<tr>
<td>z43/fem</td>
<td>0.96</td>
<td>0.99</td>
<td>1.02</td>
<td>0.97</td>
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### 10° 9

<table>
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<th>ezz</th>
<th>eil</th>
<th>match</th>
</tr>
</thead>
<tbody>
<tr>
<td>femur</td>
<td>8.3730</td>
<td>1.8219</td>
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<tr>
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</tr>
<tr>
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<td>0.99</td>
<td>0.99</td>
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### 0° 10

<table>
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<th>ezz</th>
<th>eil</th>
<th>match</th>
</tr>
</thead>
<tbody>
<tr>
<td>femur</td>
<td>6.3064</td>
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</tr>
<tr>
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<td>1.00</td>
<td>0.94</td>
<td>0.76</td>
<td>0.58</td>
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</table>
the values for the maximum stem strains in a femur, at the node locations for the maximum stem strains in a femur. The testing devices here were variations on the birchwood testing device, which was chosen as the best testing device from the parametric study. The variations were in fixture radius (rf), stem тестer interface friction (u), or material (birch or poplar). The tensile strains in the x, y and z directions are designated as exx, eyy and ezz. The interlaminar shear strain is designated as eil.

<table>
<thead>
<tr>
<th></th>
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<th>eyy</th>
<th>ezz</th>
<th>eil</th>
</tr>
</thead>
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<td></td>
<td></td>
<td></td>
</tr>
<tr>
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<td>0.87</td>
<td>0.88</td>
<td>0.77</td>
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<tr>
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<td>0.84</td>
<td>0.88</td>
<td>0.73</td>
</tr>
<tr>
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<td>0.86</td>
<td>0.88</td>
<td>0.74</td>
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</tbody>
</table>

<table>
<thead>
<tr>
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<th>eyy</th>
<th>ezz</th>
<th>eil</th>
</tr>
</thead>
<tbody>
<tr>
<td>birch</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>rf=35 mm</td>
<td>-</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
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<td>0.89</td>
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<td>0.88</td>
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</tr>
<tr>
<td>u=0.0</td>
<td>0.89</td>
<td>0.88</td>
<td>0.88</td>
<td>0.82</td>
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</table>

<table>
<thead>
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<th>eyy</th>
<th>ezz</th>
<th>eil</th>
</tr>
</thead>
<tbody>
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<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>poplar</td>
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<td>0.87</td>
<td>0.80</td>
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<table>
<thead>
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<th>ezz</th>
<th>eil</th>
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<tbody>
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<tr>
<td>poplar</td>
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<td>0.82</td>
<td>0.88</td>
<td>0.79</td>
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</table>
Table 3.10. Maximum stem strains for a composite stem in a human femur and in testing devices with proximal potting. The testing devices here were variations on the birchwood testing device, which was chosen as the best testing device from the parametric study. The variations were in fixture radius (rf), stem/tester interface friction (u), or material (birch or poplar). The tensile strains in the x, y and z directions are designated as exx, eyy and ezz. The interlamellar shear strain is designated as eil. The node locations were usually different between the maximum stem strains eyy and eil in the testing devices and in the femur. The stem strains in the testing devices at the node locations for the maximum stem strain in the femur were recorded in the columns marked “per max femur.” The ranking of how far the stem strains were below the maximum stem strains in the testing device were recorded in the columns marked “rank.” Any ranking other than one indicated a mismatch between node locations of maximum stem strains in the tester and maximum stem strains in the femur.

<table>
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3.4. Discussion

3.4.1. Simplified Models

3.4.1.1. Two-Dimensional Plus Sideplates Stem-Tester Model

This model showed that the testing device affected the stress magnitudes of a femoral stem in the testing device. However, except for changing from proximal to distal potting, the variations in the testing device studied did not affect the stress patterns much. For all testing device variations studied with this model, the maximum von Mises stem stress was located where the femoral stem entered the testing device, for both proximal and distal potting.

3.4.1.2. Axisymmetric Stem-Tester Model

Most of the variations in the testing device had very little effect on either the stress magnitudes or the stress patterns on the femoral stem. Again, changing from proximal to distal potting had a large effect. Changing from a nonrigid to a rigid tester at distal potting also had a large effect. The von Mises stem stresses damped out over a shorter distance as testing device rigidity increased. This was also seen with a beam-on-elastic-foundation model (Fig. 2.4.c).

There was much less increase in the maximum von Mises stem stress between a nonrigid and rigid testing device with proximal potting, than there was with distal potting, as seen in Table 3.11:
<table>
<thead>
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<th>Potting Device</th>
<th>Testing Device</th>
</tr>
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<tr>
<td>Proximal</td>
<td>269</td>
</tr>
<tr>
<td>Distal</td>
<td>281</td>
</tr>
<tr>
<td>Nonrigid</td>
<td>274</td>
</tr>
<tr>
<td>Rigid</td>
<td>452</td>
</tr>
</tbody>
</table>

Table 3.11. Maximum von Mises stem stresses (MPa) with proximal or distal potting, and nonrigid or rigid testing devices.

For this model and loading, the stresses on a proximally potted stem were less sensitive to a change in testing device rigidity than were the stresses on a distally potted stem. This could indicate that the design of a proximally potted testing device is less critical to the resulting stress state on the stem, than is the design of a distally potted testing device.

3.4.1.3. Axisymmetric Stem-Bone Model

The transversely isotropic stem decreased the interfacial shear strengths (IFSS's) relative to an isotropic stem in this model. A major problem with low-modulus stems is the increase in the proximal IFSS, which can lead to micromotion of the stem [35, 67, 74]. This result indicated that a low-modulus anisotropic femoral stem may decrease this in vivo problem from what would be expected with a low-modulus isotropic femoral stem.

The anisotropy, as well as the stiffness, of the femoral stem, affected stresses on the stem/bone system. Additional changes to the layup or material design of the composite stem would also be likely to affect the interfacial shear stress. This indicated an area for optimization or compromise with composite stems — designing a composite which has a low modulus yet reasonable interfacial shear stresses. These results also suggested that for finite element
analysis of composite stems, modelling the composite as a low-modulus isotropic material may be too much of a material simplification.

3.4.2. Full 3-D Models

3.4.2.1. Comparisons to Baseline Tester Values

Variations in the femoral stem testing device parameters affect the stem strains. Results from femoral stems tested in different testing devices may or may not be comparable, even if obvious parameters such as load angle, load magnitude and embedding height were the same.

The criteria used to determine the effects of changing testing parameters were the maximum stem strains exx, eyy, ezz and eil (interlaminar shear strain). Based on these criteria, varying the fixture radius had no effect above the proximal potting level, but did affect the maximum stem eil, which was just below the proximal potting level. Varying the cement modulus only affected maximum stem ezz and eil. The location of maximum stem eil changed from above to below the proximal potting level; the maximum stem eil in the neck changed little with cement modulus. Varying the stem/cement interface friction affected maximum stem eyy, ezz and eil. The location of maximum stem eil changed from below to above the proximal potting level; the maximum stem eil above the proximal potting level hardly changed with interface friction. For proximal potting, many of the maximum stem strains damped out at larger parameter values, indicating less sensitivity of the stem strain to those testing
device parameters; however, not all of the maximum stem strains damped out.

The load angle 0/10 caused the highest maximum stem strains, and the load angle 10/0 caused the lowest maximum stem strains. This demonstrated the severity of out-of-plane or torsional loads on the femoral stem, and was consistent with the literature on the testing of metallic stems [9-11] and uncemented stems [32,33]. The different load angles resulted in maximum stem strains having different magnitudes and node locations, indicating that the different load angles on the stem would result in different failure locations, magnitudes, and modes.

The embedding height at which the crossover occurred varied with the load angle and strain components. For instance, for the load angle gait, the crossover between maximum stem strain in the neck and maximum stem strain at the embedding height occurred at different embedding heights for maximum eyy and eil, indicating that the embedding height could affect the location and type of failure seen. For the load angle 0/10, the maximum interlaminar shear strain was located in the neck, until the lowest embedding height studied. By analogy, the amount of proximal resorption of a femur into which a composite stem is implanted could also affect the location and type of failure seen.

Decreased embedding heights resulted in increased stem strains, both in the neck and in the stem overall. As the embedding height decreased, the maximum stem strains increased, and the location of
the maximum stem strain crossed over from the neck to the embedding
height (Fig. 26). These results have been seen with metallic stems
[7,8]. If a femoral stem were tested at a lower embedding height,
either to raise the stresses in the neck or to cause highest stresses
at the embedding height, the objective might not be fulfilled. Neck
failures are seen in femoral stems; mechanically testing a stem at
too low an embedding height would mask neck failure, since failure
would occur at the embedding height instead of in the neck. At the
embedding height specified in ISO 7206/3 and 7206/4 for the testing
of femoral stems, all of the maximum composite stem strains, except
for the x-strain, occurred at the embedding height.

A slight decrease in embedding height from full proximal
potting changed magnitudes and locations of the maximum stem strains.
This indicated that even a slight decrease from full proximal potting
affected the stem strains.

3.4.2.2. Comparisons to Femur Values

The maximum stem strains in the testing devices were either
lower than or about equal to the maximum stem strains in the femur,
at the node locations for the maximum stem strains in the femur. As
the tester parameters were increased, the strain values were fairly
level or soon leveled off, except for maximum stem eyy. Maximum stem
exx was close between the testing device and femur values; for all
testing device parameters except angle and embedding height, maximum
stem exx stayed between 100% and 97% of the femur value. Maximum
stem eyy decreased as the tester parameters were increased, except
for the birchwood testing device (eyy increased). Maximum stem ezz leveled off to 99% of the femur value when testing device modulus was increased to 200 GPa. Maximum stem eil decreased as the tester parameters were increased, except for the birchwood testing device (eil increased), and for varying fixture radius (eil did not change).

The maximum stem strains eyy and eil in the testing devices were always worse with regards to matching the stem strains in the femur, than were exx and ezz. The maximum stem eyy and eil would lead to composite interlaminar failure, by normal and shear strain, respectively. Interlaminar failure is one of the failure modes for thick composites, and has been observed in simplified composite stems [40]. The testing devices studied here could suppress these interlaminar failures in a composite stem, relative to if the stem were in a femur. The worse match of maximum stem eyy and eil also indicated that interlaminar failures would be suppressed, relative to the in-plane tensile failures indicated by maximum stem exx and ezz. Additionally, the locations for maximum stem eyy and eil were predicted to be different for the stem in the femur and the stem in the testing devices, except for maximum stem eil in the testing devices with cement modulus of 0.5 or 1 GPa. The mismatches between the maximum stem strain in the tester and the stem strain at the node location for the maximum stem strain in the femur were indicated by the ranks (Table 3.5). For maximum stem eyy and eil, the mismatch increased as the tester parameters increased, except for the birchwood testing device. The ranks for maximum stem eil were better
than the ranks for maximum stem eyy. In fact, a remeshing of the birchwood testing device caused the maximum stem eil location to be the same as for the stem in the femur.

There was no clear load angle which resulted in the best match between the composite stem in the tester and the composite stem in the femur. Part of the reason was that for all load angles the node locations for the maximum stem strains in the tester and in the femur were different.

The best choice of testing device from this study was a birchwood testing device. This testing device supported the stem up to its neck, had an outer radius of 45 mm, and had a stem/tester interface friction of $\mu = 0.4$. Birchwood is an orthotropic material; it was considerably less stiff in the transverse and radial (x and y) directions than it was in the longitudinal (z) direction [82]. This orthotropy caused it to better simulate a femur. The stem strains were slightly improved by decreasing the tester outer radius, decreasing the interface friction, and changing the tester material to yellow poplar, a hardwood with a lower longitudinal modulus.
Fig. 3.26. Contour plots of composite stem strains in the z-direction, with proximal and distal potting. The scale was the same for both plots. The decreased embedding height caused the maximum stem strain to increase, and caused the location of the maximum stem strain to cross over from the neck to the embedding height. Similar results were seen with the other stem strains, except the location of the maximum strain in the x-direction remained in the neck for all embedding heights.

Scale (in 1E-3):  
- Red: 11.5 -> 11.5
- Orange: 7.47 -> 7.47
- Yellow-orange: 3.43 -> 3.43
- Yellow: -0.62 -> -0.62
- Green: -4.66 -> -4.66
- Lt. green: -8.71 -> -8.71
- Lt. blue: -12.8 -> -12.8
- Blue: -16.8 -> -16.8
- Dk. blue: -> -16.8
4. Finite Element Analysis - Effects on Testing Device Viability

4.1. Introduction

The objective of this study was to use finite element analysis (FEA) to parametrically examine the viability of a mechanical testing device for femoral stems. The testing device viability was determined by considering the stresses on the bone cement or tester material, and by considering the stem/cement interface motion.

4.2. Methods

The finite element models used were full 3-D models of a composite or metallic stem implanted in a cylindrical testing device. The baseline model was described in Section 3.2.2 (Fig. 3.9).

Parameters varied within the testing device model were applied load magnitude (l), fixture outer radius (rf), cement modulus (Ec), testing device modulus (Et), stem/cement interface friction (µ), applied load angle (a), and embedding height (z). The variations are listed below and are discussed in Section 3.2.2.1. The variations in boldface were those used in the baseline model; except when indicated otherwise, when one parameter was varied, all other parameters remained the same as for the baseline model.

1 - load magnitude (X BW) 2, 4, 6, 8
rf - fixture outer radius (mm) 45, 50, 55, 60
Ec - cement modulus (GPa) 0.5, 1, 3, 5, 10, 20
Et - testing device modulus (GPa) 3, 15.2, 20, 70, 100, 200
µ - interface friction 0.0, 0.2, 0.4, 0.8, 1.2, 1.6, 2.0
da - load angle designations gait, 10/0, 10/9, 0/10
z - embedding height designations z43, z39, z33, z29, z27, z25, z23

*orthotropic material
For evaluating the effects on testing device viability, with a composite stem, all testing device variations. For evaluating the effects on the bone cement or testing device material, with a metallic stem, the testing device variations studied were the baseline model, rf = 60 mm, Ec = 0.5 and 20 GPa, Et = 3, 15.2, and 200 GPa, μ = 0.0 and 2.0, and embedding height z23 (distal potting).

For the cement elements, the values generated were the principal stresses and the von Mises stresses. These stresses have been used previously for studying bone cement in finite element models [79,84,85]. These stresses were compared to strengths for bone cement, taken as representative values of those found in the literature [86,87]. The maximum tensile principal stress was compared to the uniaxial tensile strength of bone cement, 35 MPa. This is the maximum principal stress criterion, which has been used for brittle materials in tension that do not fail by yielding [88]. The maximum von Mises stress was compared to the shear strength of bone cement times the square root of three, or 40 MPa x 3, which calculates to 69 MPa. This is the von Mises or maximum octahedral shear stress criterion [88].

For increasing the cement modulus, the cement strength was recalculated, by assuming that the cement was reinforced with randomly oriented short glass fibers. The theoretical change in cement strength was calculated using planar reinforced composite theory [34,89,90]. The length-to-distance ratio of the fibers was taken as 100.
For varying the testing device modulus, there was no cement layer since the testing device was all one material. Results from the testing device element layer closest to the stem were compared to the results from the cement layer in the other models.

A beam-on-elastic-foundation (BEF) model of an implanted femoral stem system, which included a cement layer [61,64], was studied for comparison to the 3-D FEA model cement results for varying fixture radius, cement modulus, and stem modulus. This BEF model was different from the model studied in Section 2.

The interface values generated were the pressure and motion components. The interface values calculated were the shear stress and motion.
4.3. Results

4.3.1. Effects on Bone Cement

The maximum cement stresses were linear ($r^2 > 0.999$) over the load range studied. This was in spite of the nonlinearity of the full finite element model.

As the outer fixture radius increased, the maximum cement stresses decreased (Fig. 4.1). For both failure criteria studied, the maximum cement stresses were below the static strengths: 35 MPa for tensile strength and 69 MPa for von Mises strength.

As the cement modulus ($E_c$) increased, the maximum cement stresses increased (Fig. 4.2). This increase was steepest at the lower $E_c$. The theoretical cement strengths at higher $E_c$ were calculated assuming short glass fiber reinforcement. For tensile failure, the stress factor, defined as strength divided by maximum stress, increased with increasing $E_c$ (Fig. 4.3). For von Mises failure, the stress factor stayed about the same.

As the testing device modulus ($E_t$) increased, the maximum stresses in the testing device element layer closest to the stem increased (Fig. 4.4). This increase was steepest at the lower $E_t$. The orthotropic birchwood testing device, however, resulted in stresses lower than those which would be expected from an isotropic testing device with $E_t = 15.2$.

As the stem/cement interface friction ($\mu$) increased, the maximum tensile principal cement stress decreased until $\mu = 0.4$, then increased slightly, and the von Mises cement stress decreased (Fig.
The largest change in cement stresses was below $\mu = 0.4$. The maximum tensile principal stress at $\mu = 0.0$, 32.15 MPa, was very close to the tensile strength of 35 MPa.

The load angle 0/10 resulted in the highest maximum cement stresses, at all embedding heights (Figs. 4.6, 4.7). The next highest maximum tensile principal cement stresses were at load angles 10/9, gait, then 10/0, except at the highest embedding height. At the highest embedding height, the order of the last two load angles was reversed, but the values were very close. The load angles for the next highest maximum von Mises cement stresses at the highest embedding height were 10/9, 10/0, then gait. The load angles with the next highest maximum von Mises cement stresses at the four lowest embedding heights were gait, 10/9, then 10/0.

As the embedding height decreased, the maximum tensile principal cement stresses increased (Fig. 4.6). Cement failure was predicted to occur below different embedding heights for different load angles: the second highest embedding height for load angle 0/10, the third highest embedding height for load angle 10/9, the fifth highest (and third lowest) embedding height for load angle gait, and the sixth highest (second lowest) embedding height for load angle 10/0. For load angle 0/10, at the two highest embedding heights, the maximum tensile principal cement stresses were very close to the cement strength; at lower embedding heights, the cement stresses exceeded the strength. The maximum tensile principal cement stresses for load angles gait and 10/0 were close until the three lowest
embedding heights. The least change in cement stress with embedding height occurred with load angle 10/0.

As the embedding height decreased, the maximum von Mises cement stresses also increased (Fig. 4.7). Cement failure was predicted to occur below the second highest embedding height for load angle 0/10, and around the fourth highest embedding height for the other load angles. For load angle 0/10, at the two highest embedding heights, the maximum von Mises cement stresses were very close to the cement strength; at lower embedding heights, the cement stresses exceeded the strength. The maximum von Mises cement stresses for the load angles gait and 10/9 were close until the lowest embedding level. The least change in cement stress with embedding height occurred with the load angle 10/0.

At the higher stem modulus of 200 GPa, representing a metallic stem, the maximum cement stresses were about half those for the composite stem (Table 4.1). The relative influence of the tester parameters on the maximum cement stresses were close for both stem materials, except for the influence of a high testing device modulus and a low embedding height (Table 4.2). The high testing device modulus had more influence on the tester stresses with the metallic stem, and the low embedding height had more influence on the cement stresses with the composite stem.

The BEF model predicted that as fixture radius increased, the maximum proximal cement loading decreased (Fig. 4.8). This decrease was more rapid at the lower fixture radii. The BEF model also
predicted that as cement modulus increased, the maximum proximal cement loading increased, with a steeper change at the lower cement modulii (Fig. 4.9). The BEF model also predicted that as stem modulus increased, the maximum proximal cement loading decreased, with a steeper change at the lower stem modulii (Fig. 4.10).
Fig. 4.1. Effects of fixture outer radius on the maximum cement stresses in composite stem testing device finite element model. Fixture inner radius was 10 mm less than the fixture outer radius. The maximum cement stresses decreased as the fixture outer radius increased.
Fig. 4.2. Effects of cement modulus on the maximum cement stresses in composite stem testing device finite element model. The maximum cement stresses increased as the cement modulus increased.
Cement Stress Factors, Vary E(cement)

Fig. 4.3. Stress factors (strength divided by maximum stress) for bone cement in composite stem testing device finite element model, with variation in the cement modulus by adding short glass fiber reinforcement. The tensile stress factor increased as the cement modulus increased, and the von Mises stress factor stayed about the same.
Fig. 4.4. Effects of testing device modulus on the maximum testing device stresses in composite stem testing device finite element model. The testing device at 15.2 GPa was birchwood, an orthotropic material; 15.2 GPa was the longitudinal modulus. The transverse modulus was 0.762 GPa and the radial modulus was 1.19 GPa. The stresses were for the testing device element layer closest to the composite stem. The maximum cement stresses increased as the testing device modulus increased.

$\Delta$ = maximum tensile stress in birchwood
$\Theta$ = maximum von Mises stress in birchwood
Fig. 4.5. Effects of stem/cement interface friction on the maximum cement stresses in composite stem testing device finite element model. The maximum tensile principal stress first decreased, then increased slightly, and the maximum von Mises stress decreased as the friction increased.
Fig. 4.6. Effects of embedding height and load angle on the maximum tensile principal cement stresses in composite stem testing device finite element model. The horizontal line indicates the cement tensile strength (35 MPa). The load angle 0/10 resulted in the highest maximum cement stresses.
Fig. 4.7. Effects of embedding height and load angle on the maximum von Mises bone cement stresses in composite stem testing device finite element model. The horizontal line indicates the cement von Mises strength (69 MPa). The load angle 0/10 resulted in the highest maximum cement stresses.
Fig. 4.6. Effects of fixture outer radius on the maximum proximal cement loading in femoral stem testing device beam-on-elastic-foundation model. Fixture inner radius was 10 mm less than the fixture outer radius. The maximum proximal cement loading decreased as the fixture outer radius increased.
Fig. 4.9. Effects of cement modulus on the maximum proximal cement loading in femoral stem testing device beam-on-elastic-foundation model. The maximum proximal cement loading increased as the cement modulus increased.
Fig. 4.10. Effects of stem modulus on the maximum proximal cement loading in femoral stem testing device beam-on-elastic-foundation model. The maximum proximal cement loading decreased as the stem modulus increased.
Table 4.1. Maximum cement stresses in femoral stem testing device. The testing device modulus of 15.2 GPa was for birchwood, an orthotropic material; 15.2 GPa was the longitudinal modulus. The transverse modulus was 0.762 GPa and the radial modulus was 1.19 GPa.

<table>
<thead>
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<th>fixture radius (mm)</th>
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<th>von Mises (MPa)</th>
</tr>
</thead>
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<td>composite stem</td>
<td>metallic stem</td>
</tr>
<tr>
<td>45</td>
<td>17.47</td>
<td>10.63</td>
</tr>
<tr>
<td>60</td>
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<td>cement modulus (GPa)</td>
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<tr>
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<td>9.84</td>
<td>5.77</td>
</tr>
<tr>
<td>3</td>
<td>17.47</td>
<td>10.63</td>
</tr>
<tr>
<td>20</td>
<td>37.43</td>
<td>27.41</td>
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<tr>
<td>testing device modulus (GPa)</td>
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<td></td>
</tr>
<tr>
<td>3</td>
<td>20.01</td>
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</tr>
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<td>15.2 (E_L)</td>
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</tr>
<tr>
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<td>64.54</td>
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</tr>
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<td>10.63</td>
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<td>2.0</td>
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<td>12.20</td>
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<tr>
<td>embedding height</td>
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<td>proximal</td>
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<td>distal</td>
<td>91.90</td>
<td>41.46</td>
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Table 4.2. Influence of parameters on maximum cement stresses in femoral stem testing device. The denominator was the value for the baseline tester, except for the parameter of testing device modulus. The testing device modulus of 15.2 GPa was for birchwood, an orthotropic material; 15.2 GPa was the longitudinal modulus. The transverse modulus was 0.762 GPa and the radial modulus was 1.19 GPa.

<table>
<thead>
<tr>
<th></th>
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<th></th>
<th>von Mises (MPa)</th>
<th></th>
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<td></td>
<td>composite</td>
<td>metallic</td>
<td>composite</td>
<td>metallic</td>
</tr>
<tr>
<td>fixture radius</td>
<td>stem</td>
<td>stem</td>
<td>stem</td>
<td>stem</td>
</tr>
<tr>
<td>60 mm / 45 mm</td>
<td>0.78</td>
<td>0.79</td>
<td>0.66</td>
<td>0.68</td>
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<tr>
<td>cement modulus</td>
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<tr>
<td>0.5 GPa / 3 GPa</td>
<td>0.56</td>
<td>0.54</td>
<td>0.52</td>
<td>0.60</td>
</tr>
<tr>
<td>20 GPa / 3 GPa</td>
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<tr>
<td>15.2 GPa / 3 GPa</td>
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<td>0.83</td>
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<td>1.20</td>
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<td>5.44</td>
<td>4.00</td>
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<td>1.15</td>
<td>0.99</td>
<td>0.91</td>
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<tr>
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<td>distal/proximal</td>
<td>5.26</td>
<td>3.90</td>
<td>5.35</td>
<td>3.78</td>
</tr>
</tbody>
</table>
4.3.2. Effects on Interface

The maximum interface pressure, shear stress, and motion were linear \( (r^2 > 0.995) \) over the load range studied (Fig. 4.11). The maximum interface separation increased abruptly at higher than 6 X body weight (BW).

As the outer fixture radius increased, the maximum interface pressure and shear strength increased, and the maximum interface separation and motion hardly changed (Fig. 4.12). The interface motion increased slightly, but then leveled off.

As the cement modulus (Ec) increased, the maximum interface pressure and shear strength increased, and the maximum interface separation and motion decreased (Fig. 4.13). The interface shear strength leveled off at higher Ec. The interface separation and motion dropped off sharply between Ec = 0.5 GPa and Ec = 3 GPa, then leveled off at higher Ec. The interface values changed less at higher Ec.

As the testing device modulus (Et) increased, the maximum interface pressure and shear stress increased, and the maximum interface separation and motion decreased (Fig. 4.14), except for the birchwood testing device. The interface values started to level off at higher Et. The birchwood, with a longitudinal modulus of 15.2 GPa, had maximum interface pressure and shear stress close to those for Et = 3 GPa, and less than those that would be expected for an isotropic testing device with Et = 15.2. The maximum interface
separation and motion for the birchwood testing device were about
twice the expected isotropic values.

As the stem/cement interface friction (μ) increased, the
maximum interface pressure decreased, the maximum interfacial shear
stress increased, and the maximum interface separation and motion
decreased (Fig. 4.15). At μ > 0.8, everything but interfacial shear
stress was leveling off; interfacial shear stress was still
increasing, but the slope was less.

As the embedding height decreased, all the maximum interface
values eventually increased (Fig. 4.16). The changes in maximum
interface pressure and shear stress with embedding height were almost
identical; the values first decreased, then increased after the
fourth highest embedding height. The load angle 0/10 resulted in the
highest maximum interface pressure and shear stress, except the
lowest, where load angle gait had the highest values. The load angle
0/10 resulted in the highest maximum interface separation at all
embedding levels, and the highest maximum interface motion at all
embedding levels except the lowest, where load angle gait had the
highest value. The maximum interface separation and motion for load
angle 10/0 changed the least with embedding height, but the embedding
height still had an effect.
Fig. 4.11. Effects of load magnitude on the maximum stem/cement interface pressure, shear stress, separation, and motion in composite stem testing device finite element model. All interface values except separation were linear with load.
Fig. 4.12. Effects of fixture outer radius on the maximum stem/cement interface pressure, shear stress, separation, and motion in composite stem testing device finite element model. Fixture inner radius was 10 mm less than the fixture outer radius.
Fig. 4.13. Effects of cement modulus on the maximum stem/cement interface pressure, shear stress, separation, and motion in composite stem testing device finite element model.
Fig. 4.14. Effects of testing device modulus on the maximum stem/tester interface pressure, shear stress, separation, and motion in composite stem testing device finite element model, with variation in the testing device modulus. The testing device at 15.2 GPa was birchwood, an orthotropic material; 15.2 GPa was the longitudinal modulus. The transverse modulus was 0.762 GPa and the radial modulus was 1.19 GPa.
Fig. 4.15. Effects of stem/cement interface friction on the maximum stem/cement interface pressure, shear stress, separation, and motion in composite stem testing device finite element model.
Fig. 4.16. Effects of embedding height and load angle on the maximum stem/cement interface pressure, shear stress, separation, and motion in composite stem testing device finite element model.
Max Interface Motion, Vary Height

Max Interface Separation, Vary Height

Fig. 4.16. continued
4.4. Discussion

4.4.1. Effects on Bone Cement

Higher cement stresses resulted from smaller fixture radii, smaller stem/cement interface friction, higher cement and testing device modulii, the load angle 0/10, lower embedding heights, and a lower stem modulus. These factors, therefore, could adversely affect the viability of the bone cement or testing device during mechanical testing of the stem.

A BEF model also predicted higher cement stresses at smaller fixture radii, higher cement modulii, and smaller stem modulii. The increase in cement stresses with increasing cement modulus concurred with stress sharing; as a material in a construct becomes stiffer, it takes on more of the applied load. Similarly, as the femoral stem became more flexible, it took on less of the applied load, and the rest of the construct, which included the cement, took on more of the applied load.

In previous finite element studies of bone cement stresses, the cement modulus and stem modulus were varied in implanted femoral stem models [77,79]. As in the present study, a lower stem modulus and a higher cement modulus resulted in higher cement stresses.

In another finite element study of bone cement stresses, the bond strengths of both the stem/cement and cement/bone interfaces were varied between perfect bonding and no bonding in a simplified implanted femoral stem model [85]. The maximum von Mises cement stress was highest when both interfaces had no bonding, and lowest
when both interfaces had perfect bonding. This was consistent with the current study; the maximum von Mises cement stress was highest at zero stem/cement interface friction, and decreased as the interface friction increased. In the other study, the maximum principal cement stress was lowest when both interfaces were partially but not completely debonded, and was highest when both interfaces had perfect bonding. In the current study, the maximum principal cement stress was lowest at an intermediate stem/cement interface friction ($\mu = 0.4$), but the maximum principal cement stress was highest at zero stem/cement interface friction. This difference could be due to the differences between the two models; the earlier model had only transverse loading, a simplified geometry, and two interfaces.

As the cement modulus was increased, by adding short glass fibers, the stress factor for tensile cement failure increased. Although the cement took on more stress at higher modulii, the theoretical strength increase in the cement at higher modulii outstripped the stress increase. Similarly, although the testing device layer closest to the stem took on more stress at higher testing device modulii, the higher strength of higher modulus materials may be sufficient to allow these testing devices to withstand the additional stress. For instance, for delrin at $E_t = 3$ GPa, the tensile stress factor was 3.5; for cobalt-chrome at $E_t = 200$ GPa, the tensile stress factor was 13.6. The von Mises stress factor for cement failure stayed relatively constant. However, the theoretical calculations for shear strength, on which von Mises
strength is based, were not as straightforward as were the calculations for tensile strength. Three numbers were calculated, and the most conservative value was chosen.

The load angle 0/10 resulted in the highest maximum cement stresses, while the load angle 10/0 resulted in the lowest maximum cement stresses (except at the lowest embedding height). Overall, these load angles also resulted in the highest and lowest stem strains. This further indicated the severity of the load angle 0/10 for the testing of composite stems; the cement as well as the stem was highly stressed.

The embedding height at which cement failure was predicted to occur depended on the load angle. For load angle 0/10, full proximal potting, and unreinforced cement, the cement stresses were very near the strengths, for just static loading. A testing configuration for composite stems in fatigue, developed by Gavins et al., had a load angle of 0/10, proximal potting, and a load magnitude of 6675 N, which was higher than the 4671 N studied here [23]. In place of bone cement, however, was polyurethane reinforced with randomly oriented short fibers. Additionally, the embedding thickness used was well over the baseline cement thickness studied here. Both of these modifications would increase the viability of the embedding compound; the fiber reinforcement by increasing strength, and the increased thickness by decreasing stress.

For some of the testing configurations studied, the safety factors for the bone cement were rather low. At the baseline load of
6 X body weight, the stress factor for tensile cement failure with μ = 0.0 was 1.09. For the load angle 0/10, the safety factors for tensile and von Mises cement failures at the highest embedding height were 1.09 and 1.11 respectively; these values decreased at lower embedding heights, with failure predicted below the second highest embedding height. Fatigue testing at the same load magnitude and testing configuration would lower the safety factors even further, since the fatigue strength of a material is less than its static strength [84]. For a set angle, load, and embedding height, the stress factor for tensile cement failure can be increased by increasing the thickness of the bone cement, increasing the stem/cement interface friction, or by reinforcing the cement. For a set angle, load, and embedding height, the stress factor for von Mises cement failure can be increased by increasing the thickness of the bone cement, or by increasing the stem/cement interface friction.

Changing from the metallic stem to the composite stem increased the maximum cement stresses, by around twofold. This indicates that a testing device which can withstand the stresses resulting from a metallic stem may not be able to withstand the stresses resulting from a composite stem, especially in fatigue. A straingage study by Beals et al. showed that with stems embedded in composite artificial femurs, a decrease in stem modulus, by changing from a cobalt-chrome stem to an aluminum stem, resulted in an increase in the strains in the cement mantle [91]. As for the present study, there was concern that the cement fatigue strength might be exceeded for low-modulus
stems. Changing from a low testing device modulus to a high testing device modulus had more effect on the testing device stresses when the metallic stem was used. The highest testing device modulus was well above the composite stem modulus, but equal to the metallic stem modulus. For the composite stem, the effect of increasing testing device modulus would level off sooner, since the increasing testing device modulus soon exceeded the stem modulus, and the testing device could soon be considered rigid, relative to the composite stem. For the metallic stem, increasing the testing device modulus would still have an effect, since the testing device modulus only increased to the metallic stem modulus, and the testing device could not be considered rigid, relative to the metallic stem. Changing from a high to low embedding height had more effect on the cement stresses when the composite stem was used. The composite stem, being more flexible, would displace more at the lower embedding height.

For the same load angle and magnitude, the maximum cement stresses for a distal-potted metallic stem were considerably higher than the maximum cement stresses for a proximal-potted composite stem (Table 4.1). The maximum cement stresses for a distal-potted metallic stem were well above the cement strengths. The decrease in cement stresses from switching from a composite to a metallic stem did not offset the increase in cement stresses from switching from proximal to distal potting. This indicated that stems tested at lower embedding heights may require a lower load magnitude in order
to keep the cement from failing. Other testing device variables, of course, can also affect the cement viability.

4.4.2. Effects on Interface

The viability of the stem/cement or stem/tester interface was evaluated using maximum interface motion. The higher the maximum interface motion, the more the rubbing and degradation at the interface during fatigue loading, and the lower the interface viability. The maximum interface motion was lowered by decreasing fixture outer radius, and by increasing cement modulus, isotropic testing device modulus, stem/cement interface friction, and embedding height. The change in maximum interface motion with fixture outer radius was not large. The highest maximum interface motions at proximal potting were for cement modulii less than 3 GPa, stem/cement interface frictions less than 0.4, and the birchwood testing device.

Rubin et al. varied the stem/bone interface friction between 0.2 and 1.2, in a 3-D FEA model of a noncemented titanium stem in bone [83]. All interface values - pressure, shear stress, separation and motion - were sensitive to the interface friction. As for the present study, the interface separation and motion leveled off at higher interface friction, and increased exponentially at lower interface friction.

The load angle 0/10 resulted in the highest maximum interface motion at all embedding heights except the lowest, where load angle gait had the highest value. The load angle 10/0 resulted in the
lowest maximum interface motion at all embedding heights except the second highest, where it was equal to the value for load angle gait.
5. Finite Element Analysis - Summary and Conclusions

5.1. Summary

The composite stem testing devices studied here resulted in maximum stem strains which were lower than or about equal to the maximum stem strains in the femur. The testing devices were worse at inducing the correct maximum stem eyy and interlaminar shear strain than at inducing the correct maximum stem exx and ezz. The testing device which best simulated the strains of the composite stem in a femur was a single material, birchwood, rather than an outer fixture and an embedding compound. This material was orthotropic.

The viability of the testing device must also be considered when designing a testing device for femoral stems. Testing device viability was measured by evaluating maximum cement stresses and maximum interface motion; lower values indicated higher testing device viability.

A measure which improves stem strains may decrease testing device viability, while a measure which improves testing device viability may change the stem strains. For instance, improving the stem strains by decreasing interface friction reduced the testing device viability; both the maximum cement stresses and interface motion increased. Improving the stem strains by decreasing fixture outer radius decreased the maximum interface motion, but increased the maximum cement stresses. Improving the stem strains by decreasing cement modulus increased the interface motion and decreased the safety factor for cement failure; although decreasing
cement modulus decreased the maximum cement stresses, it also decreased the cement strength. Improving the safety factor of the cement by increasing fixture outer radius, interface friction, or cement modulus, affected the stem stresses. Improving the safety factor of the cement by increasing friction or cement modulus also decreased the maximum interface motion; improving the safety factor of the cement by increasing fixture outer radius only slightly increased the maximum interface motion.

Although the birchwood testing device considerably improved the maximum stem eye and interlaminar shear strain, this testing device also had high maximum interface motion. Modifying the birchwood testing device to further improve these strains also affected testing device viability. As indicated by the parametric study, decreasing fixture outer radius or interface friction increased the maximum testing device tensile stress. Decreasing the longitudinal modulus of an orthotropic testing device, by changing from birchwood to yellow poplar, decreased the maximum testing device tensile stress, but also decreased the strength; yellow poplar is weaker than birchwood [82].

5.2. Conclusions

The stresses and strains of a polymer composite femoral stem were affected by design changes to its testing device. The material of the stem also affected the stem and testing device. The simplified finite element models were effective in showing that stresses on the femoral stem were affected by design changes to its
testing device, and by changes in modulus and anisotropy of the femoral stem. Some of the results from the simplified models, however, were counter to those found with the full 3-D model. The full 3-D finite element model showed that design changes to the testing device affected not only the composite stem, but the testing device viability as well. Of the testing devices studied, birchwood, an orthotropic material, best simulated a femur, as far as the maximum stem strains were concerned.

5.2.1. Limitations of Study

The most important limitation in this study was that finite elements are, by definition, an approximation [68]. Results obtained with FEA are normally not exact results, especially with complex models. However, results obtained with FEA can indicate relative effects as parameters within the model are varied [78], and were used as such here.

Other limitations were in the simplifications for the full 3-D finite element model. The geometry of the femur consisted of cylindrical and elliptical cross-sections only. The bone properties were inhomogeneous, but isotropic. The composite stem had a quasi-isotropic layup instead of a more realistic, and more complicated, layup. The composite was also modelled as a homogeneous material instead of a layered material.

In finite element analysis, interfaces which are not fully bonded are difficult to model. The actual coefficient of friction between a noncemented femoral stem and the human femur is unknown
The interference fit between the femoral stem and the femur, which is necessary for reasonable finite element results [81], will in all likelihood not be the interference fit found in vivo [92]. Very small interference fits between the femoral stem and the femur in vivo will result in considerable stress on the femur [93].

Only one femoral stem geometric design was studied here. There may not be one optimum mechanical testing device for all composite stems. A testing device which works for one stem design might not work for another, such as a testing device for a straight stem versus a testing device for a curved stem. A testing device which works for one composite design might not work for another. A testing device for a pressfit composite stem would be expected to be different from a testing device for a composite stem which would be anchored by porous ingrowth.

5.2.2. Future Studies

More studies are necessary to design a composite stem testing device. The cylindrical geometry of the testing device, while widely used, may not be appropriate; other testing device geometries and configurations should be investigated. A FEA program with thick composite analysis capability would indicate initial failure in the composite stem. The viability of the birchwood testing device, or other hardwood testing devices, should be more precisely determined, using orthotropic strengths within a FEA program. The parametric studies could be repeated for different composite stem designs, or even for metallic stems at distal potting. Experimental stress
analysis of the stem/tester system would allow verification of the finite element results.
6. Other Studies

6.1. Introduction

Three shorter studies were done using the BEF and FEA models described earlier. The first study was to parametrically examine different configurations and variables of the mechanical testing device studied earlier, but with a metallic femoral stem. These results were compared to the results with a composite stem. The second study was to investigate the effects of stem anisotropy for the composite stem implanted in a femur. The third study was to see which of the simpler BEF and FEA models gave results that followed the same trends as the full 3-D finite element model.

6.2. Methods

6.2.1. Composite vs. Metallic Stem Strains

The finite element models used were full 3-D models of a metallic stem (E = 200 GPa) implanted in a cylindrical testing device. The baseline testing device model was described in Section 3.2.2 (Fig. 3.9).

Parameters varied within the testing device model were fixture outer radius (rf), cement modulus (Ec), testing device modulus (Et), stem/cement interface friction (μ), and embedding height (z). The variations are listed below and were discussed in Section 3.2.2.1. The variations in boldface were those used in the baseline model; when one parameter was varied, all other parameters remained the same as for the baseline model.
rf - fixture outer radius (mm)  45, 60
Ec - cement modulus (GPa)    0.5, 3, 20
Et - testing device modulus (GPa)  3, 15.2, 200
μ - interface friction    0.0, 0.4, 2.0
z - embedding height designations z43, z23

*orthotropic material

The load magnitude was 6 X BW, and the load angle was the baseline load angle, a load angle found during horizontal walking. Embedding height z43 was full proximal potting, and embedding height z23 was distal potting.

6.2.2. Effects of Stem Anisotropy

The finite element models used were full 3-D models of composite stems implanted in a human femur. This model was described in Section 3.2.2 (Fig. 3.8). The load magnitude was 6 X BW. The load angle was 10° in the medial-lateral plane and 0° in the anterior-posterior plane, which meant there was no torsional or out-of-plane loading. This load angle was chosen for better comparison to the axisymmetric models, which could not include torsional loading.

The effects of the anisotropy of the composite stem material on maximum stem strains, cancellous bone stresses, and interface values were evaluated. The orthotropic composite stem was compared to an isotropic stem and a transversely isotropic stem having engineering constants based on the orthotropic material.

6.2.3. Comparisons Between Models

The effects of testing device and femoral stem variables on the stem, cement, and interface were compared between the three
simplified models - beam-on-elastic-foundation (BEF), 2-D plus sideplates FEA, and axisymmetric FEA - and the full 3-D finite element model. Not all testing device and stem variables were studied with each model.

6.3. Results

6.3.1. Composite vs. Metallic Stem Strains

The maximum metallic stem strains exx, eyy, and ezz were lower than the maximum composite stem strains (Table 6.1). The node locations for maximum stem exx and eyy were the same between the stem in the femur and the stem in the testing devices, but were different for maximum stem ezz.

The variations of the testing device parameters fixture radius (rf), cement modulus (Ec), stem/cement interface friction (µ), and embedding height (z) had an equal or less of an effect on the metallic stem than on the composite stem, except for maximum stem ezz when varying Ec and µ (Table 5.2). Varying Ec resulted in more change in maximum stem ezz for the metallic stem; additionally, this change was in the opposite direction than for the composite stem, and the node location changed as well. Varying µ resulted in more change in maximum ezz for the metallic stem; additionally, this change was in the opposite direction than for the composite stem. Some of the stem strains were affected little by testing device parameters at proximal potting, regardless of the stem material.
6.3.2. Effects of Stem Anisotropy

The anisotropy of the composite stem material affected the maximum stem strains, cancellous bone stresses, and interface values (Table 5.3). The transversely isotropic stem resulted in the highest maximum stem strains, cancellous bone stresses, and interface values. These values were closer between the isotropic stem and the orthotropic stem than between either stem and the transversely isotropic stem.

6.3.3. Comparisons Between Models

6.3.3.1. Effects on Stem

All four models predicted that as the embedding height of the femoral stem decreased, the stem stresses or strains increased (Figs. 2.5.b, 3.12, 3.17, 3.18; Tables 3.1, 3.2). They also all predicted that at low embedding heights, maximum stem stresses or strains were located around the embedding height, rather than in the neck. The BEF, axisymmetric, and full 3-D models predicted that decreasing the embedding height also caused the stem stresses or strains in the neck to increase.

The axisymmetric model predicted a decrease in maximum stem von Mises stress as the cement thickness increased (Table 3.2). The full 3-D model predicted a decrease or no change in maximum stem strains in the neck as the cement thickness increased, but an increase in maximum stem interlaminar shear strain, which occurred below the neck (Figs. 3.14, 3.15).
The BEF model predicted that as the testing device modulus decreased, the moment on the stem neck did not change, and the moment on the stem shaft increased (Fig. 2.4.b,c). The full 3-D model predicted that as the testing device modulus decreased, maximum stem ezz increased and the other stem strains decreased, but all of the strains soon leveled off with an increase in testing device modulus (Fig. 3.21).

6.3.3.2. Effects on Bone Cement

A BEF model developed by Huiskes and Chao [61,64] predicted that maximum proximal cement loading increased with decreased fixture radius, increased cement modulus, and decreased stem modulus (Figs. 4.8 - 4.10). Similarly, the full 3-D model predicted that maximum cement stresses increased with decreased fixture radius, increased cement modulus, and decreased stem modulus (Figs. 4.1, 4.2; Table 4.1). For both models, the change in cement loading or stress was more rapid at the lower fixture radii, cement modulii, and stem modulii.

6.3.3.3. Effects on Interface

The BEF and full 3-D models both predicted that as the embedding height decreased from proximal to distal potting, the interfacial shear stress (IFSS) increased (Figs. 2.5.c, 4.16). They also both predicted an increase in IFSS with an increase in testing device modulus (Figs. 2.4.d, 4.14). For the axisymmetric model, the isotropic stem had the higher maximum IFSS (Table 3.4), and for the
full 3-D model, the transversely isotropic stem had the higher maximum IFSS (Table 6.3).
Table 6.1. Maximum normal stem strains for a metallic femoral stem in a human femur and in testing devices. The testing device variations were in outer fixture radius (rf), cement modulus (Ec), testing device modulus (Et), stem/cement interface friction (u), and embedding height (embed). The node locations were usually different between the maximum stem strain εzz in the testing devices and in the femur.

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<td>εzz</td>
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Table 6.2. Influence of parameters on maximum normal stem strains in femoral stem testing device. The denominator was the baseline tester. Numbers with asterisks indicate that maximum stem strain occurred at a different node location than for the baseline tester.

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<td></td>
<td></td>
</tr>
<tr>
<td>distal / proximal</td>
<td>7.94*</td>
<td>6.91*</td>
</tr>
</tbody>
</table>
Table 6.3. Effects of composite stem anisotropy on maximum a) stem strains, b) cancellous bone stresses, and c) stem/bone interface values.

### a) Stem Strains (1E-3)

<table>
<thead>
<tr>
<th>Stem Material</th>
<th>exx</th>
<th>evy</th>
<th>ezz</th>
</tr>
</thead>
<tbody>
<tr>
<td>orthotropic</td>
<td>5.3613</td>
<td>2.1355</td>
<td>2.2417</td>
</tr>
<tr>
<td>isotropic</td>
<td>5.2905</td>
<td>2.2649</td>
<td>1.5424</td>
</tr>
<tr>
<td>transversely isotropic</td>
<td>20.537</td>
<td>2.7356</td>
<td>3.4115</td>
</tr>
</tbody>
</table>

### b) Cancellous Bone Stresses (MPa)

<table>
<thead>
<tr>
<th>Stem Material</th>
<th>Maximum Principal</th>
<th>Minimum Principal</th>
<th>von Mises</th>
</tr>
</thead>
<tbody>
<tr>
<td>orthotropic</td>
<td>38.08</td>
<td>-60.94</td>
<td>75.87</td>
</tr>
<tr>
<td>isotropic</td>
<td>38.84</td>
<td>-61.26</td>
<td>72.64</td>
</tr>
<tr>
<td>transversely isotropic</td>
<td>58.57</td>
<td>-130.1</td>
<td>144.1</td>
</tr>
</tbody>
</table>

### c) Interface Values

<table>
<thead>
<tr>
<th>Stem Material</th>
<th>Pressure (MPa)</th>
<th>Shear Stress (MPa)</th>
<th>Motion (mm)</th>
<th>Separation (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>orthotropic</td>
<td>178.0</td>
<td>30.68</td>
<td>0.1257</td>
<td>0.0715</td>
</tr>
<tr>
<td>isotropic</td>
<td>154.9</td>
<td>31.03</td>
<td>0.1124</td>
<td>0.0699</td>
</tr>
<tr>
<td>transversely isotropic</td>
<td>345.6</td>
<td>138.29</td>
<td>0.3049</td>
<td>0.1340</td>
</tr>
</tbody>
</table>
6.4. Discussion

6.4.1. Composite vs. Metallic Stem Strains

The maximum normal stem strains were lower for the metallic stem than for the composite stem. This was no surprise, since the higher-modulus metallic stem would be expected to strain less under the same load.

Varying a testing device parameter did not necessarily have the same effect on the two stem materials. Increasing stem modulus or stem/cement interface friction at proximal potting decreased the maximum composite stem ezz but increased the maximum metallic stem ezz. Increasing the fixture radius at proximal potting had little effect on the maximum normal strains in either stem. Decreasing the embedding height caused more of an increase in maximum normal strains in the composite stem than in the metallic stem. Varying the embedding height from proximal to distal potting had the most effect on the stem strains.

6.4.2. Effects of Stem Anisotropy

The anisotropy of the composite stem material affected the stem strains, bone stresses, and stem/bone interface. The stresses and strains seen with the isotropic and orthotropic stems were lower than those seen with the transversely isotropic stem. Unidirectionally reinforced, transversely isotropic composites are not a good choice for composite stems, because of the complex, nonaxial loading to which femoral stems are subjected [94]. The transversely isotropic composite stem was also not good for simulating a more realistic
orthotropic composite stem. The results for the isotropic and orthotropic stems were still different from each other, although they were closer to each other than to the transversely isotropic composite stem or the metallic stem. This showed that modelling the composite stem as a low-modulus isotropic material may or may not be adequate to show the behavior of a composite stem/bone system, depending on what is being studied.

6.4.3. Comparisons Between Models

The full 3-D model was the most complex with regards to geometry, material properties, and loading; the others were less complex. The full 3-D model was the only model with torsional loading and an orthotropic stem material. It was also the only finite element model run using large-displacement analysis, and the only model with a nonbonded interface between the stem and the testing device. While the full 3-D model would be expected to give the most accurate results, the simpler models could in some cases show the effects of varying parameters in the implanted femoral stem system.

The simplified models showed the same effects of changing the embedding height as did the full 3-D model. The simplified and full 3-D models demonstrated that as the embedding height decreased, the stem stresses and neck stresses increased, the location of the maximum stem stress switched from the neck to the embedding height, and the interfacial shear stress (IFSS) increased.
A BEF model of a stem-cement-fixture system was very good at showing how the maximum cement stress varied with fixture radius, cement modulus, and stem modulus. Both this BEF model and the full 3-D model predicted that the maximum cement stress increased with decreased fixture radius, increased cement modulus, and decreased stem modulus. Both models also predicted sharper changes at lower parameter values. The increase in cement stresses with increased cement modulus or decreased stem modulus was also expected from considering stress sharing.

The simplified models were erratic in predicting changes in the stem and interface with other testing device and stem variables, when compared to the full 3-D model. Some of the stem strain components in the full 3-D model followed the same trends predicted by the simplified models, while others did not. The BEF and full 3-D models both predicted an increase in IFSS with an increase in testing device modulus. The axisymmetric and full 3-D model gave opposite results for the comparison between maximum proximal stem/bone IFSS with isotropic and transversely isotropic femoral stems. The axisymmetric model was inadequate to show the interface behavior when compared to the 3-D model, likely because of the lack of torsion in the axisymmetric model. However, both models showed that composite stem anisotropy was an important consideration in FEA.
Metallic femoral stems have a standardized, well-established protocol for in vitro mechanical testing, while polymer composite femoral stems do not. The testing device configuration most commonly used for metallic stems simulates proximal resorption of the femur, by only supporting or embedding the distal portion of the stem. This testing device is inappropriate for composite stems, since they are expected to reduce or prevent proximal resorption. The purpose of this study was to use finite element analysis (FEA) to determine the effects of changing a testing device on the composite stem, and to find a testing device for the composite stem which best simulated the composite stem in the femur.

Three simplified models used for preliminary studies of stem/tester or stem/bone systems were beam-on-elastic-foundation, two-dimensional plus sideplates FEA, and axisymmetric FEA. These models had limitations in loading, geometry, and material. However, they still gave some results equivalent to those found with the full 3-D FEA models. They were also used to help decide what to study with the full 3-D FEA models.

The effects of varying the testing device on the maximum normal and interlaminar shear strains of the composite stem were studied using full 3-D FEA models. The composite stem was symmetrical, straight-stemmed, and orthotropic. The testing devices were a cylindrical outer fixture and bone cement (as an embedding compound), or a cylindrical single-material testing device. Testing device
variables studied were load magnitude and angle, fixture outer radius, cement modulus, testing device modulus, stem/cement interface friction, and embedding height. The maximum stem strains were linear with applied load. The load angle of 10° out-of-plane caused the highest stem strains, while the load angle of 10° in-plane caused the lowest stem strains. The maximum stem strains were affected to various degrees by geometric and material variations to the testing device at the highest embedding height. Decreasing the embedding height, even slightly, caused stem strains to increase.

The embedding height at which the location of the maximum stem strains switched from the neck to the embedding height, varied with the strain component and load angle. Distal potting of a femoral stem may be inappropriate when trying to induce neck failure, since failure may occur at the embedding height instead of in the neck.

The maximum normal and interlaminar shear strains of the composite stem in the testing device were either lower than or about equal to the maximum strains of the composite stem in the femur. The maximum stem y-strain and interlaminar shear strain, which lead to composite interlaminar failure, were the most underestimated by the testing devices. Additionally, the locations for these strains were different between the stem in the femur and the stem in the testing devices.

Birchwood, an orthotropic hardwood, was chosen as the best testing device from the parametric study. This testing device greatly improved the maximum stem y-strain and interlaminar shear
strain, relative to the other testing devices studied. Decreasing the testing device radius, stem/tester interface friction, or longitudinal modulus of the hardwood, slightly improved the stem strains.

The viability of the testing device, as measured by stresses on the bone cement or testing device material and by interface motion, was also affected by changes to the testing device and stem material (Tables 7.1, 7.2). Changing from a metallic stem to a composite stem increased the bone cement stresses by about twofold, indicating that a testing device strong enough for a metal stem may not be strong enough for a composite stem. Some testing device configurations had cement stresses that were close to or above static cement strengths; dynamic loading would even worsen the cement viability. For a set load magnitude, angle, and embedding height, the stresses on the bone cement or testing device material decreased as fixture outer radius and interface friction increased. Increasing the cement modulus by adding short-fiber reinforcement increased the cement stresses, but also increased the cement strengths, resulting in an increased safety factor at higher cement modulii. For a set load magnitude, angle, and embedding height, the interfacial motion decreased as fixture outer radius decreased, and as cement modulus, testing device modulus and interface friction decreased. Although the birchwood testing device was the best testing device as far as the stem was concerned, it had high interfacial motion. Further improvements to the birchwood testing device, with regards to improving the maximum stem
strains, decreased the viability of the testing device. The load angle of 10° out-of-plane resulted in the worst testing device viability, while the load angle of 10° in-plane resulted in the best testing device viability.

Other stem materials briefly studied were metal, low-modulus isotropic, and low-modulus transversely isotropic. The strains on the metallic stem and composite stem were not always affected in the same way by changes to the testing device. The transversely isotropic stem was not a good simplification of the orthotropic composite stem. The low-modulus isotropic stem was better, but still gave different results from the orthotropic composite stem.
Table 7.1. Femoral stem testing device parameters which increase maximum cement stresses.

<table>
<thead>
<tr>
<th>Increased Cement Stresses</th>
</tr>
</thead>
<tbody>
<tr>
<td>Decreased fixture radius</td>
</tr>
<tr>
<td>Decreased interface friction</td>
</tr>
<tr>
<td>Increased cement modulus</td>
</tr>
<tr>
<td>Increased testing device modulus</td>
</tr>
<tr>
<td>Load angle 10° out-of-plane</td>
</tr>
<tr>
<td>Decreased embedding height</td>
</tr>
<tr>
<td>Decreased femoral stem modulus</td>
</tr>
</tbody>
</table>

Table 7.2. Femoral stem testing device parameters which increase maximum interface motion.

<table>
<thead>
<tr>
<th>Increased interface motion</th>
</tr>
</thead>
<tbody>
<tr>
<td>Increased fixture radius</td>
</tr>
<tr>
<td>Decreased cement modulus</td>
</tr>
<tr>
<td>Decreased testing device modulus</td>
</tr>
<tr>
<td>Decreased interface friction</td>
</tr>
<tr>
<td>Increased embedding height</td>
</tr>
</tbody>
</table>
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