Comparison of human and canine external femoral morphologies in the context of total hip replacement

Roy D. Bloebaum[1,2,*] Doug T. Ota,[1,2] John G. Skedros,[3] and John P. Mantas[1]

[1]*Bone and Joint Research Laboratories (151F), V.A. Medical Center, 500 Foothill Boulevard, Salt Lake City, Utah 84148;* [2]*Division of Orthopedic Surgery, University of Utah School of Medicine, 50 N. Medical Drive, Salt Lake City, Utah 84132; and* [3]*LAC/University of Southern California Medical Center, Department of Orthopedics (RM 3900), Box 699, General Hospital, 1200 North State Street, Los Angeles, California 90033*

The canine is frequently used as a model for human hip arthroplasty research. In order to better understand the appropriateness of the canine as a model for human total hip replacement studies, the external morphology of canine and human femurs were examined and compared. Several differences were found between canine and human femora, including angular measurements, anterior bow, and femoral head position relative to the femoral diaphysis. In addition, the human femur was noted to undergo age-related changes in several of the measured parameters. The canine femur did not exhibit any age-related changes in the measured parameters. This study suggests that there are limitations to the use of the canine model in human hip arthroplasty research, and that discretion must be exercised when attempting to extrapolate results from a canine study to the human clinical condition. © 1993 John Wiley & Sons, Inc.

INTRODUCTION

The canine model remains the animal of choice for determining the efficacy and safety of various materials and designs used in total hip arthroplasty. While justification for the use of the canine model has been published in the past,[1-4] neither a careful review of the literature concerning the appropriateness of the canine model nor a comparative study between human and canine femora morphology, in the context of hip function, could be found. With the continued use of the canine hip as a model for human arthroplasty research, there is a need to examine the canine femur in order to understand the appropriateness of this model for total hip replacement (THR) research. The objective of this communication is to compare various external morphologic femoral features in a highly cursorial canine breed to the human femur in the context of hip function and age-related changes. This study also reviews previous comparative studies of the canine and human femur and discusses them in the context of research in THR.

*To whom correspondence should be addressed.

Literature review

Adverse changes in bone quality and bone mass are known to occur with alarmingly high incidence in the proximal human femur after total hip replacement. These adverse changes have been implicated as major factors in causing aseptic mechanical loosening in total hip replacement.[5-19] In an effort to develop THR designs that avoid clinical failures associated with bone resorption, mechanical loosening, and other adverse tissue reactions, a variety of animal models have been used to evaluate the response of bone tissue to various prosthetic designs. The canine is the most widely used animal model in this regard and has been considered by some investigators to be useful in predicting the ultimate human clinical success or failure associated with various prosthetic design features. These have included implant material and mechanical properties[6,20-27] optimal placement of the porous coating,[13,20,28,29] and the capacity of various types of porous coatings to allow or induce bony ingrowth for skeletal fixation.[6,18,26,29-33]

In a study supporting the use of the canine as an appropriate model for simulating loading conditions seen in the human femur, Goel et al. concluded that canine and human femora are similar

Journal of Biomedical Materials Research, Vol. 27, 1149–1159 (1993)

CCC 0021-9304/93/091149-11

in terms of various external and internal anatomical features.[2] Other investigators also supported the use of the canine as a model for human THR research since the human and canine hips have a similar load orientation,[4] share similar cortical microstructure[1] and cortical blood supply.[1,34]

Sumner and coworkers showed that various external angles in the proximal canine femur, which are important considerations in THR because they influence the shape of the prosthesis, differed substantially from those measured in a large sample of human femora.[3] Furthermore they showed that the canine has relatively thinner cortical bone since the canine medullary canal is larger than the human medullary canal relative to external femoral dimensions. Despite these differences in cross-sectional geometry and angles, Sumner et al. supported the use of the canine model and suggested that it may provide a model of accentuated bone loss in hip arthroplasty experiments.

Although these studies have addressed important considerations in the selection of the canine as a model for THR research, little attention has been specifically focused on various gross anatomical differences that could influence how the results of canine studies are extrapolated to human clinical conditions. Investigations that support the possibility that the canine and human hip endure widely different functional stressing include the documentation of marked differences in their relative limb weight distributions,[35] femoral strain magnitudes,[36,37] and locomotor patterns both with and without hip replacements.[38]

Age-related differences in canine and human bone mass distributions and geometry, and bone quality are also important considerations when evaluating the appropriateness of using the dog as an experimental model for human THR. In the human femur, age-related changes are known to occur in bone mass distributions and geometry, and bone quality.[39-45] Although the relationship of these factors to clinical failure of implanted femoral components has been recognized,[46,47] their relative importance among other factors in causing prosthetic failure is poorly understood.[48]

Age-related changes in the response of bone tissue to implanted femoral components have been examined in canines. In a study examining the interface strength between canine bone and a porous-coated femoral component, Magee et al. showed a marked difference between younger and older canines.[49] However, no studies have examined a broad age range of canine femora for age-related external geometric changes that are relevant to THR research, even though the aging human femur is known to undergo changes in various anatomical features that can substantially alter mechanical stresses across the hip.[41,42,46,50-52]

Morphometric study

Several of the external morphologic features examined in the present study have been suggested as being important in previous investigations, but have not been quantified. Sumner and co-workers stated that the canine femoral head offset length, stated in terms of femoral neck length, and normalized to bone length, is much shorter in canines than humans.[3] They suggest that this difference may account for the lack of relationship between the distribution of bone within the proximal diaphysis, measured as the major axis of the second moment of inertia, and various external angles of the canine femur. Therefore, femoral head offset length is a feature that warrants further comparative study since this parameter can have an important influence on the magnitude of bending moments imparted across the femoral neck.[41,51,53] Sumner and co-workers also stated that the canine and human femora both have an anterior bow with an apex in the mid-diaphyseal region.[3] However, we have recognized a marked difference in the relative location of maximum anterior curvature in the femora of these two species, and we hypothesize that with certain stem designs this may be an important consideration. Goel et al. examined various important external anatomical features in canine and human femora but failed to statistically examine their data.[2]

The present study attempts to clarify the potential use of the canine model in human THR by accomplishing two main objectives: the examination, quantification, and statistical comparison of various external anatomical features in canine and human femora that are felt to be important in human THR, and the analysis of the structural changes mentioned above for the aging human femur to see if they also occur in the aging canine femur.

METHODS

Forty-two left and right greyhound femora (18 male, 24 female, age range 1–10 years) and 30 left and right Caucasian human femora (21 male, 9 female, age range 17–72 years) were studied. The morphometric parameters measured in this study were anteversion angle (AV), cervico-diaphyseal angle (CD), total bone length (L), biomechanical length (L_b), head diameter in anteroposterior plane (HD_{ap}), head diameter in superioinferior plane (HD_{Si}), length from head center to lateral margin of greater trochanter (L_{ht}), length from head center to longitudinal axis of diaphysis (L_{hd}), femoral head offset length (L_{ho}), anterior bow (AB), anterior bow at point of maximal curvature (AB_{max}), and anterior bow index (AB_i).

The methods described by Ruff were adopted for establishing the anatomical axis system and global or three-dimensional orientation.[41] However, the distal supracondylar region of the canine femur does not have the distinct anterior "low" point that is evident along the anterior cortex in the supracondylar region of the human femur that was used by Ruff to define the distal point of the longitudinal axis in lateral view.[41] Therefore, in canines the distal mark on the lateral diaphysis was made on the cadaveric bone at the point just proximal to the consistently prominent epicondylar tubercle.

Biomechanical length, total length, head center, angular measurements, and bow measurements were all made on the cadaveric bone with the bone oriented. Biomechanical and total length were measured using an osteometric table configured with a sliding vernier caliper. Other length measurements were made with hand-held calipers. The head center was determined using a template with concentric circles placed over the cadaveric femoral head in the frontal plane when the femora were oriented, in accordance with Ruff.[41]

Figures 1, 2, and 3 illustrate the various direct measurements made on the specimens. Total and biomechanical bone lengths were measured parallel to the longitudinal diaphyseal axis. Biomechanical length (L_b) was measured in both species from the point where the longitudinal axis intersected the proximal femur, which occurred typically in the region where the greater trochanter merges with the base of the femoral neck, to the point along the longitudinal axis at one-half the distance between the most inferior edges if the condyles. The distance between the inferior condylar edges was averaged in this manner because of the asymmetry of the human condyles.[41] Total bone length was measured from the superior point of the head to the point one-half the distance between the most inferior edges of the condyles. In canines there was no need to average the distance between the inferior edges of the condyles since the condyles are virtually symmetrical.[54]

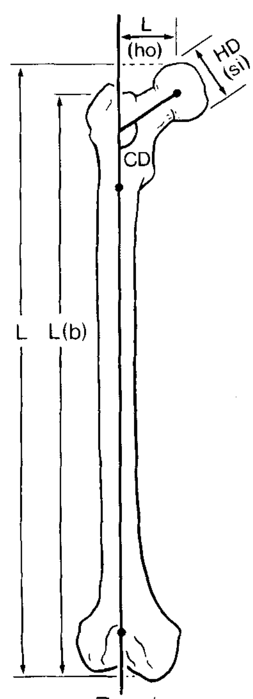

Figure 1. Morphometric parameters measured in this study for the comparison between canine and human femora. Z axis represents the longitudinal diaphyseal axis.

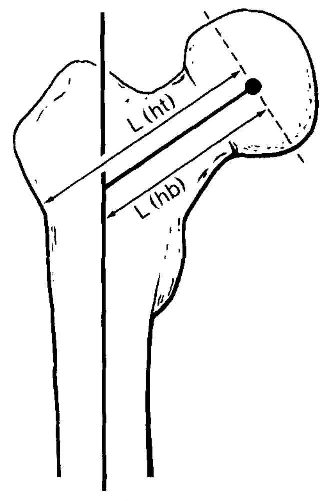

Figure 2. The length from the head center to the lateral margin of the greater trochanter [L(ht)] was measured parallel to the cervical axis. The length from the head center to the longitudinal diaphyseal axis [L(hb)] was measured parallel to the cervical axis.

The length from the head center to the lateral margin of the grater trochanter (L_{ht})] was measured parallel to the cervical axis. The length from the head center to the longitudinal diaphyseal axis (L_{hd}) was measured parallel to the cervical axis. The femoral head offset length (L_{ho}) was measured as the horizontal distance from the head center to the longitudinal diaphyseal axis, perpendicular to this axis. These three measurements were made in the plane of the anteversion angle. Head diameter was measured in the anteroposterior (HD_{ap}) and superioinferior (HD_{si}) directions. All these measurements were made directly on the specimens while oriented on the osteometric table.

The anteversion angle and the cervico-diaphyseal angle in the greyhound were the only measurements made using the biplanar radiographic techniques described by Montavon et al.[55] This method can be summarized as follows: radiographs of the femurs were taken in the anteroposterior (AP) and mediolateral (ML) planes (Figs. 4[a, b]). The X and Y coordinates used to calculate the anteversion angle were

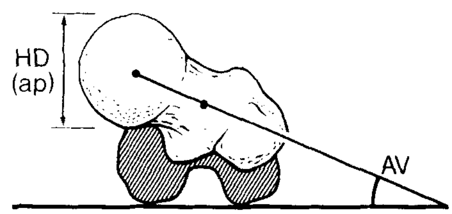

Figure 3. Illustration of the head diameter measured in the anteroposterior plane [HD(ap)] and the anteversion angle [AV]. The anteversion angle is formed by the inclination of the femoral head and neck from the transcondylar axis in the frontal plane.

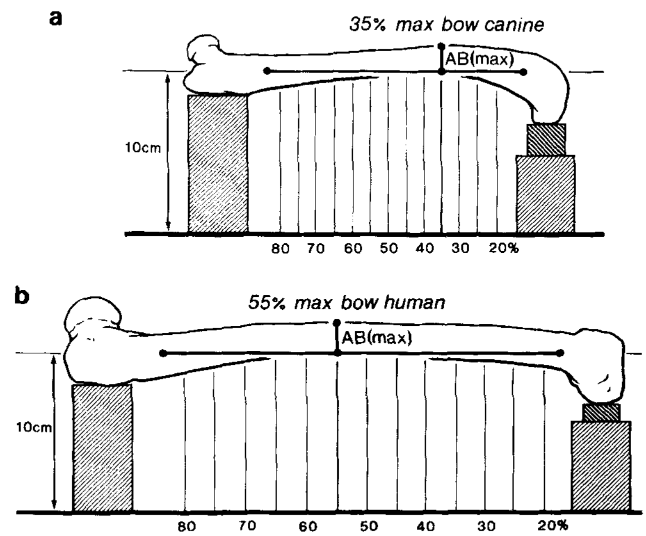

Figure 5. Anterior bow was measured from the leveling surface to the anterior cortex at intervals of 5% of the biomechanical length starting at 20% and ending at 80%. Canines (Fig. 5a) had a maximum bow at 35% of their biomechanical length. Humans (Fig. 5b) had a maximum bow at 55% of the biomechanical length.

established on the radiographs. The Y coordinate was defined as the distance from the center of the head to the longitudinal axis on the AP radiograph. The X coordinate was defined as the distance from the center of the head to the longitudinal axis on the ML radiograph. The anteversion angle was calculated from the relationship tan(AV) = X/Y. The cervicodiaphyseal angle of the canine femur was measured directly on the AP radiograph and corrected to the real angle of inclination using trigonometric relationships described by Ogata and Goldsand.[56]

Measurements for anterior bow were also made directly on the specimens while the femora were oriented on the osteometric table according to the methods described by Ruff.[41] Small holes were drilled into the proximal and distal points that define the longitudinal diaphyseal axis along the lateral aspect of the diaphysis. The femurs were then elevated above the leveling surface on which the condyles rested to a distance where the axis formed by these two holes was 10 cm above the original leveling surface (Figs. 5 [a, b]). Global anatomical orientation was also preserved while anterior bow measurements were made. Elevation of the femur placed human and canine femora at the same plane of reference, which facilitated both anterior curvature measurements and subsequent between-species comparisons. The locations at which anterior bow measurements were made were calculated as percentages of biomechanical length starting at the 20% location, which is the distal or condylar end, and proceeding proximally at 5% increments to the 80% location, which is near the base of the lesser trochanter. At each percent length, the height of the anterior cortex above the leveling source was measured. To compare the amount of anterior bow in the AP plane between the two species, a relative height or anterior bow index (AB_i) was calculated. The anterior bow index is defined as the distance from the anterior cortex at the location with the maximum bow to the leveling axis, which is 10 cm above and parallel to the leveling surface, divided by biomechanical length ($AB_i = AB_{max} - 10)/L_b$).

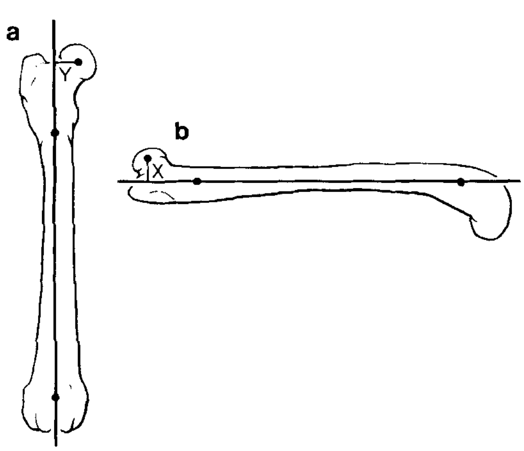

Figure 4. Biplanar method for measuring the anteversion angle in the canine. Y was measured on the anteroposterior radiograph. X was measured on the mediolateral radiograph. The anteversion angle was calculated by the relationship tan(AV) = X/Y.

Potential size- and sex-related variability in the data was reduced by dividing the remaining length measurements by the biomechanical length and multiplying by 100. This statistical approach for reducing variability by normalizing data is based on a study by Ruff that allows ratios to be compared. Ruff showed that human femoral cross-sectional geometric properties scale geometrically with bone length.[57] The same scaling relationships are thought to also occur in the canine femur.[3] Differences between parameters were evaluated statistically using a nonparametric Mann-Whitney test with statistical significance set at $P < .05$.

Age-related changes in canine and human femoral morphology were also analyzed. Human data were divided into a young group with a mean age of 26 years (range 17–36 years, n = 16) and an older group with a mean age of 64 years (range 50–89 years, n = 14). Greyhound data were also divided into a young group, with a mean age of 21 months (range 12–29 months, n = 18) and an older group with a mean age of 69 months (range 35–136 months, n = 24). Age-related comparisons were also evaluated statistically using a nonparametric Mann-Whitney test with significance established at $P < .05$

RESULTS

Table I lists the mean and standard deviation for each parameter measured or calculated in canines and humans, and P value of the comparisons between canines and humans. Normalized data are listed in Table I.

When comparing the entire samples without distinguishing age, several geometric differences between canine and human femora were identified. Humans have a smaller anteversion angle ($P < .05$) and a smaller cervico-diaphyseal angle ($P < .05$). The ratio of total bone length (L) to biomechanical length (L_b) was greater in humans than canines. The canine has a relatively longer distance from the head center to the longitudinal axis of the diaphysis (L_{hd}/L_b) ($P < .05$). No significant between-species differences were found in the length from the head to the lateral margin of the greater trochanter (L_{ht}/L_b) ($P = .07$), or in the femoral head offset length (L_{ho}/L_b) ($P = .54$).

In canines, there was no significant difference between anteroposterior head diameter (HD_{ap}/L_b) and superioinferior head diameter (HD_{si}/L_b) ($P = .97$). However, measurements of femoral head diameters in humans showed that the AP diameter was greater than the SI diameter ($P < .05$). This suggests that the human femoral head may be a spheroid that is more oblong than that of the canine femoral head.

Comparison of the anterior bow index measured at the point of maximum bow showed that the greyhound is significantly more bowed in the anteroposterior direction than humans ($P < .05$). The point of maximum bow is located more distally in dogs, at approximately 35% of the biomechanical length, than in humans where the maximum bow occurred at approximately 55% of the biomechanical length (Table II, Figs. 5 [a, b], 6).

Table III lists the means, standard deviations, and P values for the comparisons made between young and old human femora. No age-related change was found at any diaphyseal location in terms of the anterior bow indices. No age-related change was found in the anteversion angle in the human femora. However, there was a significant age-related decrease in the human cervico-diaphyseal angle of 5.8° ($P < .05$). In human femora, significant age-related increases were also found in the distance from the head to the lateral

TABLE I
The Mean, Standard Deviation (in parentheses), and P Values of the Compared Parameters between Humans and Canines. Symbols Presented as a Ratio Indicate Normalized Data. (See text for parameter abbreviations.)

Parameter	Canine	Human	P
AV	27.0° (6.3)	7.8° (6.0)	< .05
CD	139.8° (3.1)	128.0° (8.1)	< .05
L_b	291.6 mm (9.9)	431.0 mm (28.3)	< .05
L/L_b	103.8 (0.7)	106.2 (1.2)	< .05
HD_{si}/L_b	10.6 (0.5)	10.6 (0.7)	.90
HD_{ap}/L_b	10.6 (0.4)	11.0 (0.7)	< .05
L_{ht}/L_b	23.8 (0.9)	23.2 (1.6)	.07
L_{hb}/L_b	15.2 (1.5)	12.8 (1.9)	< .05
L_{ho}/L_b	9.8 (1.1)	10.0 (1.8)	.54
AB_i	6.8 (0.8)	5.5 (0.8)	< .05

margin of the greater trochanter (L_{ht}/L_b) ($P < .05$), head center to the longitudinal axis (L_{hd}/L_b) ($P < .05$), and femoral head offset length (L_{ho}/L_b) ($P < .05$).

Comparisons between the young and old greyhound groups in Table IV showed that there were no significant age-related changes in any of the parameters, including angular and length measurements, and anterior bow indices.

Table V lists the P values of the comparisons made between the entire greyhound sample to each of the young and old human groups. The canine remained significantly different ($P < .05$) from both the young and older humans with respect to anteversion angle (canine > human), cervico-diaphyseal angle (canine > human), length from the head center to the longitudinal axis (L_{hd}/L_b) (canine > human), and anterior bow index at the location of maximum bow (canine > human). When the length of the head center to the lateral margin of the greater trochanter (L_{ht}/L_b) in the human group was compared to the greyhound, it was observed that there was a significant difference between the young humans and the entire canine sample (canine > human), but no significant difference could be shown between the older human group and the entire canine sample. Comparison of femoral head offset lengths (L_{ho}/L_b) showed that there was a significant difference between the entire greyhound sample and the young and old human age groups; the younger human sample had a mean femoral head offset length that was smaller than the canine ($P < .05$), but the older human sample had a mean head

TABLE II
Anterior Bow: Average height (cm) of the Anterior Cortex above the Leveling Surface and the Standard Deviation (in parentheses) for Each Percent Length

	20%	25%	30%	35%	40%	45%	50%	55%	60%	65%	70%	75%	80%
Canine	11.20	11.25	11.42	11.48	11.45	11.41	11.22	11.25	11.18	11.11	11.03	10.99	10.92
	(0.10)	(0.10)	(0.11)	(0.14)	(0.16)	(0.16)	(0.16)	(0.16)	(0.15)	(0.15)	(0.14)	(0.19)	(0.12)
Human	12.01	12.09	12.16	12.21	12.24	12.29	12.72	12.36	12.35	12.29	12.17	11.95	11.70
	(0.20)	(0.21)	(0.23)	(0.26)	(0.30)	(0.31)	(0.34)	(0.36)	(0.36)	(0.34)	(0.30)	(0.27)	(0.26)

offset length that was greater than the canine ($P < .05$). In contrast, there was no significant difference between femoral head offset length between canine and human femora when each species was examined with no distinction made for age.

DISCUSSION

In the present study several morphological differences between human and greyhound femora were identified. Two of the most obvious differences are the cervico-diaphyseal angle and the anteversion angle. Montavon et al.[55] reported values of the mean anteversion and the mean cervico-diaphyseal angles in adult mongrel dogs as being 31.3° and 144.7° respectively, and Sumner et al.[3] reported 34.2° and 147.4°, respectively, in their series of mongrel dogs. These values are higher than those reported in the present study, where mean anteversion angle measured 27.0° and mean cervico-diaphyseal angle 139.8°. Consequently, in terms of these two angular measurements, greyhound femora differ from adult mongrel dogs. In humans, the mean anteversion angle measured 7.8° for the entire sample, with no distinction for sex made, which is similar to the mean anteversion angle of 7° in male and 10 degrees in female Caucasians reported by Hoaglund and Low.[58] This is at the lower range of normal when compared to numerous other studies cited by Clark et al.[53] In the human femora examined in the present study, the mean cervico-diaphyseal angle measured 128.0°, which is similar to the mean values reported by Goel et al.[2] and in numerous other studies reviewed by Clark et al.[53] Humans, therefore, have a femoral neck that is significantly less anteverted and a cervico-diaphyseal angle that places the hip in a relatively more varus position than seen in both mongrel dogs and greyhounds. Additionally, a few human specimens in our sample were retroverted (n = 3). Retroversion of the femoral neck was not seen in our sample of greyhounds, nor has it been previously reported in other canine studies.

The present study showed differences in anterior bow of the diaphysis between canines and humans. The point of maximum bow in the human femora occurred at approximately 55% of the biomechanical length, which is consistent with the previous findings of Walensky.[45] In contrast, the greyhound femora had the point of maximum curvature located more distally, at approximately 35% of biomechanical length. The canine femora also exhibited a greater degree of relative curvature (AB_i). These differences in curvature of the femur must be considered when

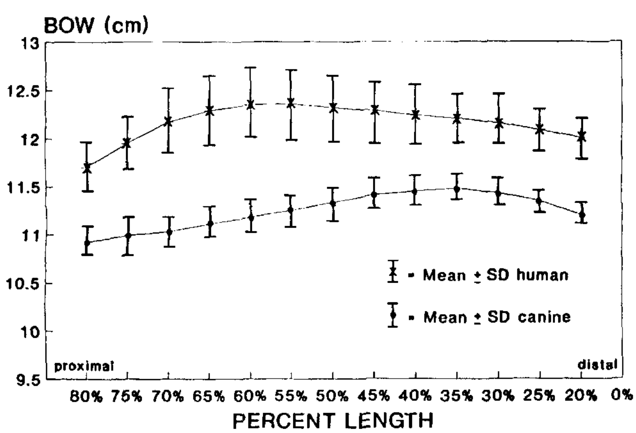

Figure 6. Plot of the means and standard deviations of the anterior bow measurements, at each percent length, for canines and humans. Humans have a maximum bow anteriorly at 55% of their biomechanical length. Canines have a maximum bow anteriorly at 35% of their biomechanical length.

TABLE III
Means, Standard Deviations (in parentheses), and *P* Values for the Age-Related Comparisons of the Human Parameters

Parameter	Young	Old	*P*
AV	8.2°	7.4°	.57
	(1.1)	(1.8)	
CD	130.7°	124.8°	< .05
	(1.9)	(2.1)	
L_{ht}/L_b	22.5	24.0	< .05
	(0.4)	(0.3)	
L_{ho}/L_b	8.9	11.3	< .05
	(0.4)	(0.3)	
L_{hb}/L_b	11.8	13.8	< .05
	(0.5)	(0.3)	
AB_i	5.5	5.5	.56
	(0.2)	(0.2)	

TABLE IV
Means, Standard Deviations (in parentheses), and *P* Values for the Age-Related Comparisons of the Canine Parameters.

Parameter	Young (12–29 months)	Old (35–136 months)	P
AV	25.0° (6.1)	28.5° (6.1)	.08
CD	139.7° (3.8)	139.8° (2.6)	.83
L_{ht}/L_b	23.6 (1.1)	24.0 (0.8)	.18
L_{ho}/L_b	9.6 (1.1)	9.9 (1.0)	.15
L_{hb}/L_b	14.9 (1.3)	15.4 (1.7)	.67
AB_i	6.8 (1.0)	6.8 (0.7)	.92

TABLE V
***P* Values of the Comparisons between Canines and Younger Humans and between Canines and Older Humans**

Parameter	Canines vs. Younger Humans	Canines vs. Older Humans
AV	< .05	< .05
CD	< .05	< .05
L_{ht}/L_b	< .05	.55
L_{hb}/L_b	< .05	< .05
L_{ho}/L_b	< .05	< .05
AB_i	< .05	< .05

interpreting bone adaptive responses that are thought to be due to dynamic loading conditions on the femoral stem. For example, whether the femoral stem is straight or curved may bear relevance to how the implant is stressed in humans and canines. An anatomical curved stem, designed to improve canal filling and cortical contact in humans that is modelled experimentally in the relatively straight proximal femoral diaphysis of the greyhound, may encounter a markedly different degree of cortical contact that may even approximate three point fixation. This difference must be considered when interpreting the results of canine studies since the stiffness of the femoral stem and surrounding bone, and its fit into the proximal femur are critical design features in influencing stress shielding.[59]

Bending moments across the proximal femur and femoral head offset length are influenced by the angulation of the neck in three-dimensional space.[41,53] In order to maintain the original balance of abductor and joint reaction forces so that hip loading conditions can remain in a physiologic range after THR, endoprosthetic designs must restore these geometric relationships.[51] If age-related changes in these geometric features occur in the proximal human femur, but not in the canine femur, this would represent potential disparities in the loading conditions imparted across the implanted hip components and surrounding bone. Results of the comprehensive osteometric anthropological study reported by Ruff[41] showed that age-related changes occurred in the human femur in both femoral neck anteversion and cervico-diaphyseal angles. Ruff[41] showed that, with age, these changes were much more conspicuous in females, where the femoral anteversion decreased by 6.8° ($P < .01$) and the cervico-diaphyseal angle decreased by 1.1° ($P > .05$). Although these changes may appear small, the associated change in femoral stressing could substantially decrease the bending moment, and hence protect the femoral neck by decreasing tensile stress.[41] Results of the present study showed that in human femora there was an age-related decrease in the mean cervico-diaphyseal angle of 5.8° ($P < .05$), but no significant change in the anteversion angle ($P = .57$). Additional age-related changes identified in our sample of human femora included increases in the length from the head center to the greater trochanter (L_{ht}/L_b) ($P < .05$), femoral head offset length (L_{ho}/L_b) ($P < .05$), and the length from the head center to the longitudinal diaphyseal axis (L_{hd}/L_b) ($P < .05$). These differences can be attributed to the apparent decrease in cervico-diaphyseal angle which alters the head position relative to the femoral diaphysis.[53] Varus and valgus orientation of the proximal femur is believe to influence the bending moment across the hip by altering in the femoral head offset length and hence the effective length of the lever arm formed by the quasi-cantilevered construction of the proximal femur.[51,53,60–62] Since the typical recipient of a total hip replacement is an elderly patient, age-related changes in femoral morphology must be considered since these changes can have substantial impact on local loading conditions. These age-related changes in humans and the absence of these change in the canine sample should be considered when attempting to extrapolate the results from long-term canine THR studies to the longevity of an artificial human hip.

It is interesting that the mean femoral head offset length of the canines in this study was relatively greater than that of the young human group, but less than that of the older human group. It has been suggested that this age-related change in femoral head offset length may represent an adaptation in the human hip to changes in predominate stress patterns.[41] Whether or not a similar structural change occurs in the canine proximal femur is not known. Although no change in femoral head offset was seen in this study of canine femora, it is suggested that dogs that exceed the age of our sample may be needed to address this issue.

Since geometric and allometric changes can occur in limb bones during growth, it is important to address the following question: when do growth-related changes cease to occur in the limb skeleton of greyhound dogs? Smith[63] and Smith and Allcock[64] have provided data that partially answer this question by showing that all femoral physes in greyhounds fuse between 11 and 12 months of age. Additionally, these investigators showed that virtually all growth plates had fused by 14 months of age. Torzilli et al.[65] also showed that structural properties of canine limb bones do not change substantially after 7 months of age. Based on these observations it is suggested that researchers should avoid the use of greyhounds less than 16 months of age in order to reduce the possibility of growth-related effects on both bone geometric and mechanical properties. In terms of the geometric parameters examined in the present study, including angular and length measurements, the greyhounds from 1–3 years of age did not differ significantly from those from 8–10 years of age. It is important to note that even though external anatomical differences do not vary with age in our sample, various internal features dealing with bone and related tissue constitution or physiology may still be undergoing changes that could yield differential responses to devices implanted in a broad age range of animals.

Magee et al.[49] have examined the effects of age on the development of bone-implant interface strength in canine femora. In their experiment, they observed that at 6 weeks the young greyhounds attained a higher bone-implant interface strength than the older group, and that at 7 weeks the older group had not yet achieved the interface strength the young group achieved at 6 weeks. In their view, the bone response rate to an implanted femoral component is influenced by more than gross morphology and loading conditions, but also by the influence of the aging process on bone tissue's ability to remodel. They suggested that studies done on canines without regard to age may be inappropriate for extrapolation to the relationship of bone response to prosthetic design in a typical elderly human recipient. Other authors have suggested that the faster remodeling rate in the canine bone may explain why reproducible bone ingrowth is seen in this model and rarely in human bone.[66,67]

Poss et al.[68] have summarized the variables of biologic and prosthetic design features that influence the performance of total joint prostheses. Bobyn et al.[6] listed important variables that can affect bone modeling and remodeling around a femoral component. These variables can be implant- and animal- related, and include implant material and modulus of elasticity, location of the porous coating, size of the stem relative to the size of the femur, the amount of rotation or angulation of the stem within the femur, apparent density of bone interfacing with the implant, biological variability in the animals' response to surgery and their tendency to form and resorb bone, degree and rate of loading, and period of implantation. Since clinical and experimental studies of total hip replacement have shown that a close geometric fit between the femoral component and supporting bone is essential for durable implant fixation,[51,69,70] it can be argued that in this situation it is of cardinal importance that the anatomy of the bone be precisely known.[71] The present study provides an anatomical database for future studies using the greyhound femur to model total hip replacement designs. The appropriateness of an animal model, however, can not simply be established on morphologic criteria alone. Investigators must also be cognizant of other differences between canines and humans when attempting to extrapolate results from canine studies to human clinical situations.

By virtue of having a quadrupedal gait and digitigrade stance, it is immediately apparent that dogs have limb weight distributions that are different from those occurring in human limbs during their bipedal plantigrade gait. To state it plainly, it requires little intuition to recognize that dogs walk on four limbs and humans on two. Thus, it seems, that loads placed on the limb bones and joints must be different in these species. Recent studies have examined these differences and have attempted to quantify the relevant loads across the hip in order to clarify the role of the canine hip in modeling THR designs for human use. Loading conditions at the hip are influenced not only by proximal femoral morphologies, but also by the global functional morphologies of musculoskeletal and ligamentous elements and associated locomotor patterns of the animal. Dogan et al.[72] quantified intersegmental forces, moments of inertia, and various kinematic parameters of the canine hind limb in order to identify parameters predictive of restoration of normal gait following total hip replacement. They found no alteration in kinematic parameters as defined by time in stance phase and by the position, velocity, and acceleration of the body segments. However, they did find that intersegmental joint forces and moments were significantly decreased in the implanted hip but did not increase significantly in the control hip, indicating that no significant compensation was occurring. Yet in these implanted animals they observed a bilateral decrease in hip forces, indicating that there may be compensation for THR by shifting weight from the hind limbs to the fore limbs.

Hip joint reaction forces have also been quantified in the human and canine hip in order to clarify the role of the canine as an experimental model. Grelsamer et al.[38] report that the canine center of gravity is located in the vertebral column at approximately 40% of the distance from the shoulder to the pelvis. Therefore, more weight is carried by the fore limbs than the hind limbs. Page et al.[73] showed that dogs typically place at least twice as much weight on the fore limbs than on the hind limbs. Allan et al.[35] determined weight distribution in the canine hip using foot-floor reaction force measurements and showed that a hind limb receives approximately 0.35 × body weight (BW), which corresponds to a hip reaction force of 0.9 × BW. In contrast, the floor-reaction force was 0.65 × BW for the fore limb, which is nearly twice that of the hind limb. In canines, Grelsamer et al.[38] estimated hip forces during the gait cycle to range from 1.66 × BW at maximum loading to 0.36 × BW when the forepaw was directly under the shoulder. In comparison, hip forces have been reported to be dramatically greater in humans, normally ranging from 2.4–6.5 × BW, according to activity level.[74–78] Even when size normalization is done, these differences still demonstrate that during normal activity much larger loads are being carried by the human hip. Furthermore, Grelsamer et al.[38] found that when canines walked, the force produced by the hip abductors was almost completely unloaded in the walking mode when the contralateral forepaw was directly under the shoulder. This decrease in hip force occurred when the dog had three paws that were simultaneously on the ground, which is common in the walking cycle. Therefore, they concluded that unless vigorously exercised at a gait that consistently keeps only two legs in a weight bearing position, dogs do not load their hip joints anywhere near the force levels typically experienced by the human hip. These species differences may be important contributing factors, in addition to differences in skeletal morphologies, to the disparities in clinical successes that occur between canines and humans, since the role of body weight and range of cyclic stress fluctuations across the hip play an important role in the fatigue life of the femoral stem.[79]

Reliance on human studies for the evaluation of the interaction between host bone and femoral prosthesis types and various prosthetic design features can be experimentally and ethically problematic. The collection of clinical data and the procurement of postmortem prosthetic retrievals can be difficult to obtain in relatively short periods of time. Even when such data or specimens are available, there are still instances when new endoprostheses must be tested for biocompatibility and longevity in biologic conditions prior to human use.[27,80] This often necessitates the use of experimental animals, particularly canines, as models for evaluating the functional adaptive and pathological response of bone and other tissues to the prosthesis. Although use of an animal model involves a departure from human biomechanical and physiological conditions, the discrepancies may be reduced by understanding both the limitations of the animal model and the specific constraints imposed by the experimental design and by the questions under consideration. Our findings, using a homogenous breed of highly cursorial canines, showed that there are significant geometric differences between greyhound and human femora. These differences suggest that marked disparities exist in loading conditions between the two species. It can be concluded that discretion must be used when attempting to extrapolate results from studies using this breed of canine to the human clinical condition.

The authors recognize the support of the DVA medical research funds, Salt Lake City, UT. They are also indebted to Ms. Gwenevere Shaw for her excellent attention to detail in the preparation of the manuscript and Ms. Cathy Sanderson for her expert technical experience.

References

1. F. Eitel, F. Klapp, W. Jacobson, and L. Schweiberer, "Bone regeneration in animals and in man. A contribution to understanding the relative value of animal experiments to human pathophysiology." *Arch. Orthop. Traumat. Surg.*, **99**, 59 (1981).
2. V. K. Goel, H. Drinker, M. M. Panjabi, and A. Strongwater, "Selection of an animal model for implant fixation studies: Anatomical aspects," *Yale J. Biol. Med.*, **55**, 113 (1982).
3. D. R. J. Sumner, T. C. Devlin, D. Winkelman and T. M. Turner, "The geometry of the adult canine proximal femur, *J. Orthrop. Res.*, **8**, 671, (1990).
4. G. Bergmann, J. Siraky, and A. Rohlmann, "A comparison of hip joint forces in sheep, dog and man," *J. Biomech.*, **17**, 907 (1984).
5. R. D. Beckenbaugh and D. M. Illstrup, "Total hip arthroplasty. A review of three hundred and thirty-three cases with long follow-up," *J. Bone Joint Surg.*, **60-A**, 306 (1978).
6. J. D. Bobyn, R. M. Pilliar, A. G. Binnington, and J. A. Szivek, "The effect of proximally and fully porous-coated canine hip stem design on bone modeling," *J. Orthop. Res.*, **5**, 393 (1987).
7. I. W. Brown and P. A. Ring, "Osteolytic changes in the upper femoral shaft following porous-coated hip replacement," *J. Bone Joint Surg.*, **67-B**, 218 (1985).
8. J. G. DeLee and J. Charnley, "Radiological demarcation of cemented sockets in total hip replacement," *Clin. Orthop.*, **121**, 20 (1976).
9. C. A. Engh and J. D. Bobyn, "The influence of stem size and extent of porous coating on femoral bone resorption after primary cementless hip arthroplasty," *Clin. Orthop.*, **231**, 7 (1988).
10. C. A. Engh, J. D. Bobyn, and A. H. Glassman, "Porous-coated hip replacement. The factors governing bone ingrowth, stress shielding, and clinical results," *J. Bone Joint Surg.*, **69-B**, 45 (1987).
11. T. A. Gruen, G. M. McNeice, and H. C. Amstuts, "'Modes of failure' of cemented stem-type femoral components. A radiographic analysis of loosening," *Clin. Orthop.*, **141**, 17 (1979).
12. W. H. Harris, A. L. Schiller, J.-M Scholler, R. A. Freiberg, and R. Scott, "Extensive localized bone resorption in the femur following total hip replacement," *J. Bone Joint Surg.*, **58-A**, 612 (1976).
13. A. K. Hedley, I. C. Clarke, S. C. Kozinn, I. Coster, T. Gruen, and H. C. Amstutz, "Porous ingrowth fixation of the femoral component in a canine surface replacement of the hip." *Clin. Orthop.* **163**, 300 (1982).

14. A. A. McBeath and R. N. Foltz, "Femoral component loosening after total hip arthroplasty," *Clin. Orthop.*, **141,** 66 (1979).
15. I. Oh and W. H. Harris, "Proximal strain distribution in the loaded femur" *J. Bone Joint Surg.*, **60-A,** 75 (1978).
16. T. P. Schmalzried, L. M. Kwong, M. Jasty, R. C. Sedlacek, T. C. Haire, D. O. O'Connor, C. R. Bragdon, J. M. Kabo, A. J. Maccolm, and W. H. Harris, "The mechanism of loosening of cemented acetabular components in total hip arthroplasty," *Clin. Orthrop.*, **274,** 60 (1992).
17. R. N. Stauffer, "Ten-year follow-up study of total hip replacement with particular reference to roentgenographic loosening of the components," *J. Bone Joint Surg.*, **64-A,** 983 (1982).
18. D. R. Sumner and T. M. Turner, "The effects of femoral component design features on femoral remodeling," in R. H. J. Fitzgerald, (ed.), *Non-Cemented Total Hip Arthroplasty,* New York, Raven Press, Ltd. (1988).
19. C. J. Sutherland, A. H. Wilde, L. S. Borden, and K. E. Marks, "A ten-year follow-up of one hundred consecutive Müller curved-stem total hip-replacement arthroplasties," *J. Bone Joint Surg.*, **64-A,** 970 (1982).
20. J. D. Bobyn, H. Goto, J. J. Krygier, A. H. Glassman, C. E. Brooks, and J. E. Miller, "The effect of stem stiffness on femoral bone resorption after canine porous coated hip replacement," *Trans. Orthrop. Res. Soc.*, **15,** 202 (1990).
21. J. D. Bobyn, E. S. Mortimer, A. H. Glassman, C. A. Engh, J. E. Miller, and C. E. Brooks, "Producing and avoiding stress shielding. Laboratory and clinical observations of noncemented total hip arthroplasty," *Clin. Orthrop.*, **274,** 79 (1992).
22. P.-Q. Chen, T. M. Turner, H. Ronningen, J. Galante, R. Urban, and W. Rostoker, "A canine cementless total hip prosthesis model," *Clin. Orthop.*, **176,** 24 (1983).
23. J. A. Longo, F. P. Magee, S. E. Mather, R. Yapp, J. B. Koeneman, and A. M. Weinstein, "Comparison of HA and non-HA coated carbon composite femoral stems," *Trans. Orthopaed. Res. Soc.*, **14,** 384 (1989).
24. F. P. Magee, A. M. Weinstein, J. A. Longo, J. B. Koeneman, and R. A. Yapp, "A canine composite femoral stem. An *In vivo* study," *Clin. Orthop.*, **235,** 237 (1988).
25. R. M. Pilliar, H. U. Cameron, R. P. Welsh, and A. G. Binnington, "Radiographic and morphologic studies of load-bearing porous-surfaced structured implants," *Clin. Orthrop.*, **156,** 249 (1981).
26. D. R. Sumner, T. M. Turner, R. M. Urban, and J. O. Galante, "Long-term femoral remodelling as a function of the presence, type and location of the porous coating in cementless THA," *Trans. Orthopaed. Res. Soc.*, **13,** 310 (1988).
27. D. R. Sumner, T. M. Turner, R. M. Urban, and J. O. Galante, "Experimental studies of bone remodeling in total hip arthroplasty," *Clin. Orthop.*, **276,** 83 (1992).
28. C. R. Bragdon, M. Jasty, G. Russotti, E. Cargill, T. P. Harrigan, and W. H. Harris, "Patterns of bone ingrowth, cortical porosity, and bone mass in a proximally porous coated canine femoral implant after six month implantation," *Trans. Orthop. Res. Soc.*, **14,** 378 (1989).
29. T. M. Turner, D. R. Sumner, R. M. Urvan, D. P. Rivero, and J. O. Galante, "A comparative study of porous coatings in a weight-bearing total hip-arthroplasty model," *J. Bone Joint Surg.*, **68-A,** 1396 (1986).
30. P. E. Greis, J. D. Kang, V. Silvaggio, and H. E. Rubash, "A long-term study on defect filling and bone ingrowth using a canine fiber metal total hip model," *Clin. Orthop.*, **274,** 47 (1992).
31. W. H. Harris and M. Jasty, "Bone ingrowth into porous coated canine acetabular replacements: The effect of pore size, apposition, and dislocation," in *The Hip,* R. H. J. Fitzgerald (ed.), St. Louis, Missouri, C. V. Mosby Co., (1985).
32. M. Jasty, C. R. Bragdon, S. Schutzer, H. Rubash, T. Haire, and W. H. Harris, "Bone ingrowth into porous coated canine total hip replacements. Quantification by backscattered scanning electron microscopy and image analysis," *Scan. Microsc.*, **3,** 1051 (1989).
33. T. M. Turner, D. R. Sumner, R. M. Urban, and J. O. Galante, "Cortical remodeling and bone ingrowth in proximal and full-length porous-coated canine femoral stems," *Trans. Orthopaed. Res. Soc.*, **13,** 309 (1988).
34. F. W. Rhinelander, C. L. Nelson, R. D. Stewart, and C. L. Stewart, "Experimental reaming of the proximal femur and acrylic cement implantation," *Clin. Orthop.*, **141,** 74 (1979).
35. C. Allan, M. Jasty, T. P. Harrigan, C. Bragdon, E. Cargill, R. W. Mann, M. Murphy, and W. H. Harris, "A comparison of the joint reaction forces measured with an instrumented prosthesis and calculated from gait analysis," *Trans. Orthopaed Res. Soc.*, **14,** 226 (1989).
36. M. Jasty, A. Page, R. Henshaw, C. Bragdon, E. Cargill, and W. H. Harris, "Long duration measurement *in vivo* strain distribution in the canine femur," *Trans. Orthopaed. Res. Soc.*, **13,** 236 (1988).
37. A. Page, M. Jasty, D. O. O'Connor, R. Henshaw, and W. H. Harris, "Strain alterations produced by cementless total hip arthroplasty in the canine. An *in vivo* biomechanical study," *Trans. Orthopaed. Res. Soc.*, **13,** 361 (1988).
38. R. Grelsamer, H. Alexander, M. Fruchtman, J. Kowalski and D. Shetty, "Three-dimensional analysis of canine hip forces," *Trans. Orthopaed. Res. Soc.*, **14,** 224 (1989).
39. R. B. Martin and P. J. Atkinson, "Age and sex-related changes in the structure and strength of the human femoral shaft," J. Biomech., **10,** 223 (1977).
40. R. Poss, "Natural factors that affect the shape and strength of the aging human femur," *Clin. Orthop.*, **274,** 194 (1992).
41. C. Ruff, "Structural changes in the lower limb bones with aging at Pecos Pueblo," Ph.D. Thesis, University of Pennsylvania (1981).
42. C. B. Ruff and W. C. Hayes, "Sex differences in age-related remodeling of the femur and tibia," *J. Orthop. Res.*, **6,** 886 (1988).
43. E. D. J. Simmons, K. P. H. Pritzker and M. D. Grynpas, "Age-related changes in the human femoral cortex," *J. Orthop. Res.*, **9,** 155 (1991).
44. R. W. Smith and R. R. Walker, "Femoral expansion in aging women: Implications for osteoporosis and fractures," *Science,* **145,** 156 (1964).
45. N. A. Walensky "A study of anterior femoral curvature in man," *Anat. Rec.*, **151,** 559 (1965).
46. A. A. Hofmann, R. W. B. Wyatt, E. P. France, G. T. Bigler, A. U. Daniels, and W. E. Hess, "Endosteal bone loss after total hip arthroplasty," *Clin. Orthop.*, **245,** 138 (1989).

47. R. Poss, P. Staehlin and M. Larson "Femoral expansion in total hip arthroplasty," *J. Arthroplasty,* **2,** 259 (1987).
48. R. Huiskes, H. Weinans, and B. van Rietbergen, "The relationship between stress shielding and bone resorption around total hip stems and the effects of flexible materials," *Clin. Orthop.,* **274,** 124 (1992).
49. F. P. Magee, J. A. Longo and A. K. Hedley, "The effect of age on the interface strength between porous coated implants and bone," *Trans. Orthopaed. Res. Soc.,* **14,** 575 (1989).
50. M.F. Ericksen, "Aging changes in the medullary cavity of the proximal femur in American blacks and whites." *Am. J. Phys. Anthrop.,* **51,** 563, (1979).
51. P. C. Noble, J. W. Alexander, L. J. Lindahl, D. T. Yew, W. M. Granberry and H. S. Tullos, "The anatomic basis of femoral component design," *Clin. Orthop.,* **235,** 148 (1988).
52. C. B. Ruff and W. C. Hayes, "Subperiosteal expansion and cortical remodeling of the human femur and tibia with aging," *Science,* **217,** 945 (1982).
53. J. M. Clark, M. A. R. Freeman and D. Witham, "The relationship of neck orientation to the shape of the proximal femur," *J. Arthroplasty,* **2,** 99 (1987).
54. V. S. de Kleer, "Development of bone," in *Bone in Clinical Orthopaedics. A Study in Comparative Osteology,* G. Sumner-Smith (ed.) Philadelphia, W. B. Saunders Co. (1982).
55. P. M. Montavon, R. B. Hohn, M. L. Olmstead and R. L. Rudy, "Inclination and anteversion angles of the femoral head and neck in the dog. Evaluation of a standard method of measurement," *Vet. Surg.,* **14,** 277 (1985).
56. K. Ogata and E. M. Goldsand, "A simple biplanar method of measuring femoral anteversion and neck-shaft angle," *J. Bone Joint Surg.,* **61-A,** 846 (1979).
57. C. B. Ruff, "Allometry between length and cross-sectional dimensions of the femur and tibia in *Homo sapiens sapiens,*" *Am. J. Phys. Anthrop.,* **65,** 347 (1984),
58. F. T. Hoaglund and W. D. Low, "Anatomy of the femoral neck and head, with comparative data from Caucasians and Hong Kong Chinese," *Clin. Orthop,* **152,** 10 (1980).
59. D. R. Sumner and J. O. Galante, "Determinants of stress shielding: Design versus materials versus interface," *Clin. Orthop.,* **274,** 202 (1992).
60. P. G. J. Maquet, *Biomechanics of the Hip: As Applied to Osteoarthritis and Related Conditions,* New York, Springer-Verlag (1985).
61. F. Pauwels, *Biomechanics of the Normal and Diseased Hip,* Berlin, Springer-Verlag (1976).
62. E. Radin, "Biomechanics of the human hip," *Clin. Orthop.,* **152,** 28 (1980).
63. R. N. Smith, "Radiological observations on the limbs of young greyhounds," *J. Small Animal Pract.,* **1,** 84 (1960).
64. R. N. Smith and J. Allcock, "Epiphyseal fusion in the greyhound," *Vet. Rec.,* **72,** 75 (1960).
65. P. A. Torzilli, K. Takebe, A. H. Burstein, and K. G. Heiple, "Structural properties of immature canine bone," *J. Biomech. Eng.,* **103,** 232 (1981).
66. R. D. Bloebaum, M. Merrell, K. Gustke, and M. Simmons, "Retrieval analysis of a hydroxyapatite-coated hip prosthesis," *Clin. Orthop.,* **267,** 97 (1991).
67. A. A. Hofmann, R. D. Bloebaum, K. N. Bachus, M. H. Rubman, and R. Plaster, "Microscopic analysis of autograft bone chips applied at the bone/implant interface of porous coated devices in human cancellous bone," *Int. Orthop.* (SIROT Prize) **16,** (1992).
68. R. Poss, P. Walker, M. Spector, D. T. Reilly, D. D. Robertson and C. B. Sledge, "Strategies for improving fixation of femoral components in total hip arthroplasty," *Clin. Orthop.,* **235,** 181 (1988).
69. J. P. Collier, M. B. Mayor, J. C. Chae, V. A. Surprenant, H. P. Surprenant, and L. A. Dauphinais, "Macroscopic and microscopic evidence of prosthetic fixation with porous-coated materials," *Clin. Orthop.,* **235,** 173 (1988).
70. A. Page, M. Jasty, C. Bragdon, K. Ito, and W. H. Harris, "Alterations in femoral and acetabular bone strains immediately following cementless total hip arthroplasty: An *in vitro* canine study," *J. Orthop. Res.,* **9,** 738 (1991).
71. P. J. Rubin, P. F. Leyvraz, J. M. Aubaniac, J. N. Argenson, P. Estève, and B. de Roguin, "The morphology of the proximal femur. A three-dimensional radiographic analysis," *J. Bone Joint Surg. (Br.),* **74-B,** 28 (1992).
72. S. Dogan, P. A. Manley, R. J. Vanderby, S. S. Kohles, L. M. Hartman, and A. A. McBeath, "Canine intersegmental hip joint forces and moments before and after cemented total hip replacement," *J. Biomech,* **24,** 397 (1991).
73. A. Page, T. P. Harrigan, C. Allen, M. Jasty, R. W. Mann, and W. H. Harris, "Analysis of the canine gait using bone pins, bilateral force plates and rigid segment motion analysis system," *Trans. Orthopaed. Res. Soc.,* **13,** 544 (1988).
74. R. D. Crowninshield, R. C. Johnston, J. G. Andrews, and R. A. Brand, "A biomechanical investigation of the human hip," *J. Biomech.,* **11,** 75 (1978).
75. V. T. Inman, "Functional aspects of the abductor muscles of the hip," *J. Bone Joint Surg.,* **29,** 607 (1947).
76. R. D. McLeish and J. Charnley, "Abduction forces in the one-legged stance," *J. Biomech.,* **3,** 191 (1970).
77. E. F. Rybicki, F. A. Simonen, and E. B. J. Weis, "On the mathematical analysis of stress in the human femur," *J. Biomech.,* **5,** 203 (1972).
78. N. Rydell, "Forces acting in the femoral head-prosthesis," *Acta Orthop. Scand.,* **37,** 88 Suppl. 1 (1966).
79. T. P. Andriacchi, J. O. Galante, T. B Belytschko, and S. Hampton, "A stress analysis of the femoral stem in total hip prostheses," *J. Bone Joint Surg.,* **58-A,** 618 (1976).
80. D. R. Sumner, T. M. Turner, R. M. Urban, and J. O. Galante, "Remodeling and ingrowth of bone at two years in a canine cementless total hip-arthroplasty model," *J. Bone Joint Surg.,* **74-A,** 239 (1992).

Received September 2, 1992
Accepted March 20, 1993