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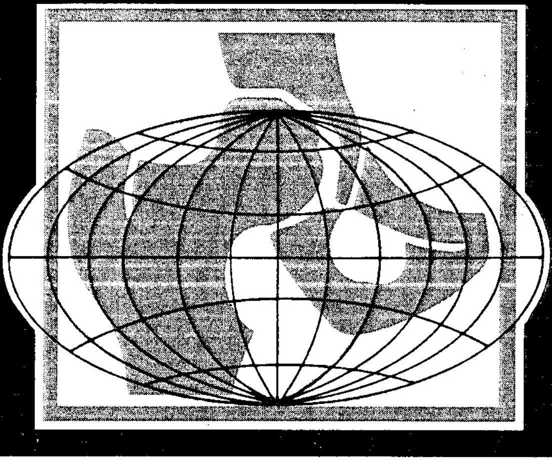

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The relationship between femoral morphotype and femoral component design

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ABSTRACT: The shape of the femoral canal is variable, much more so than most contemporary designs of femoral components would suggest or can accommodate. In the face of this variability, line-to-line or surface-to-surface contact is not expected between the cementless implant and much of the endosteal surface. It is also apparent that changes in implant design are still needed if the normal biomechanics of the hip joint are to be restored in each patient and if component fixation is to be optimized. Most cementless components aim to achieve proximal load transfer to the femoral canal. However, increasing clinical evidence suggests that distal filling of the femur is also necessary to minimize the incidence of postoperative symptoms, particularly in revision procedures. If this is indeed the case, more accommodating designs of femoral components are needed that will permit proximal and distal fitting at the femoral canal so that stable fixation may be achieved regardless of variations in bone geometry. (Hip International 1995; 5: 113-20)

KEY WORDS: Femoral component design, Anatomy of proximal femur

INTRODUCTION

A close geometric fit between the femoral component and the bone is essential for lasting implant fixation. The high incidence of aseptic loosening following cemented total hip arthroplasty (tha) emphasizes this necessity considering the direct relationship between the load capacity of the bone-cement interface and bone fastness. More lasting stability of cementless THA is also due to a better geometrical arrangement between the bone end femoral component. For this reason the success of the THA is related to the predictability of the femoral dimensions.

Despite the importance of matching the shape of the promixal femur, few studies have examined endosteal geometry in the design of the femoral prosthesis. It has been suggested that the stem design may be based on the relationship derived from periosteal geometry extrapolated to the shape of the endosteal canal (1). Others have suggested that femoral prostheses, even with a finite number of sizes, may provide contact with much of the endosteal surface of the proximal femur (2-3). Yet another theory has it that the femoral stem may be selected on the reamed diameter of the medullary canal (4).

To examine these proposals and to study the relationship between the design of the femoral stem and endosteal geometry, an anthropometric study of the proximal femur was carried out.

MATERIAL AND METHODS

Roentgenograms of 150 femora were prepared using the standard technique with views parallel and perpendicular to the femoral neck. A roentgenographic scale allowed compensation for magnification which was typically 1.5%. The population undergoing x-ray examination had an average age

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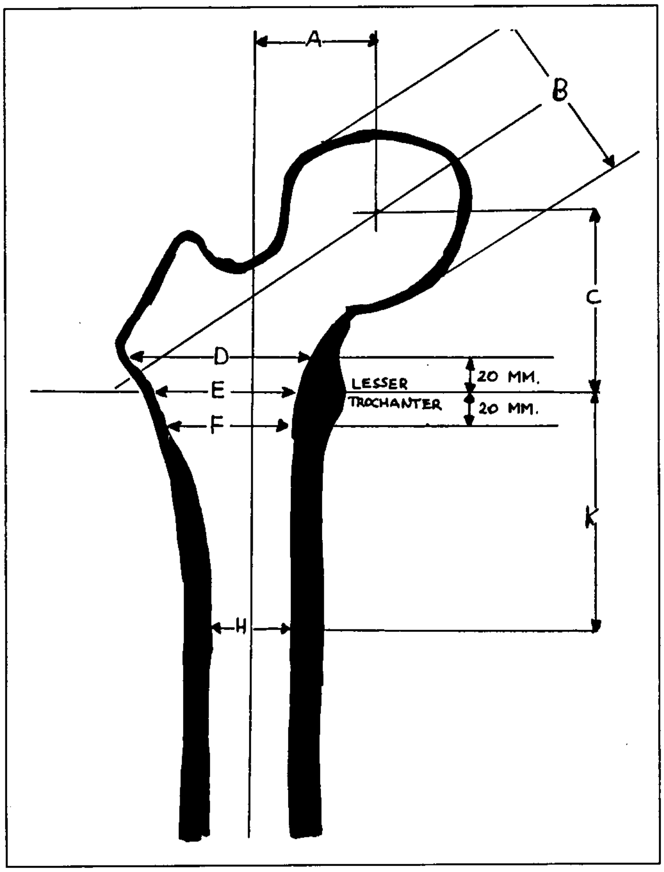


Fig. 1 - *Diagrammatic representation of the dimensions of the femur from the antero-posterior view.*

of 70.2 years, ranging from 22 to 95 years.

The periosteal and endosteal dimensions were determined for each femur by direct measurement of each roentgenogram (Fig. 1).

To permit comparison of the dimensions of bones of different sizes, horizontal and vertical reference axes were established using the geometric center of the lesser trochanter (LT) and the bisecting axis of the medullary canal at the isthmus. The mean and standard deviations were calculated for each femur dimension.

For each dimension of the proximal femur, dimensions of bones with canals, in the bottom 10% of the anatomical distribution were determined using correlation analysis. A proportional index, defined as the ratio of the 90% to 10% dimensions, was calculated to express the relative increase in each femoral dimension across the normal anatomic range.

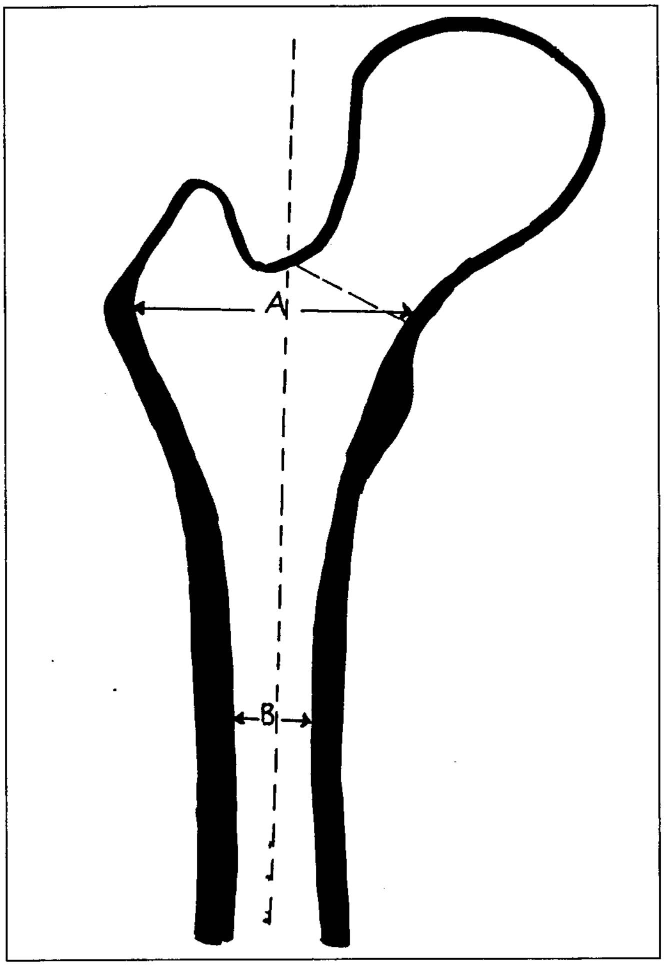


Fig. 2 - *Diagrammatic representation of the "canal flare index".*

Canal shapes were subjectively classified as normal, stove-pipe or champagne fluted. These shapes were expressed by a single geometric parameter called the canal flare index defined as the ratio of the intracortical width of the femur at a point 20 mm proximal to the lesser trochanter and at the canal isthmus (Fig. 2).

Anatomic results

Average dimensions for all of the femora examined in this study, presented in Table I, demonstrate the substantial variability in every instance.

All dimensions were significantly associated at the 5% level, with the exception of the neck-shaft angle. This variability was not purely random but the relationship between the two variables was not use-

TABLE I - VARIABLES OF FEMORAL SPECIMENS

Dimension	No.	Average	Standard deviation	Minimum	Maximum
Femoral head offset	150	43.0	6.8 mm	23.6 mm	61.0 mm
Femoral head diameter	150	46.1 mm	4.8 mm	35.0 mm	58.0 mm
Femoral head position	150	51.6 mm	7.1 mm	32.8 mm	74.3 mm
Canal width lesser troch. +20 mm	150	44.8 mm	5.9 mm	30.9 mm	59.4 mm
Canal width lesser troch	150	28.9 mm	4.2 mm	16.8 mm	41.0 mm
Canal width lesser troch. -20 mm	150	21.2 mm	3.7 mm	10.9 mm	30.1 mm
Extracortical width mediolateral	150	27.3 mm	3.5 mm	19.9 mm	35.8 mm
Isthmus width mediolateral	150	12.6 mm	2.5 mm	7.6 mm	18.9 mm
Proximal border of Isthmus	150	86.4 mm	18.1 mm	36.8 mm	193.4 mm
Distal border of Isthmus	150	143.0 mm	18.9 mm	93.2 mm	204.8 mm
Isthmus position	150	112.9 mm	16.6 mm	63.5 mm	154.8 mm
AP canal width (osteotomy level)	150	23.8 mm	3.5 mm	14.9 mm	30.9 mm
Medial diameter of femoral neck	150	17.2 mm	3.1 mm	9.9 mm	22.8 mm
Isthmus width AP	150	16.8 mm	3.9 mm	10.4 mm	26.6 mm
Neck-shaft angle	150	125.2°	6.9°	108.6°	148.6°
Femoral length	150	438.5.0 mm	30.4 mm	358.1 mm	545.1 mm
Effective neck length	150	34.9 mm	2.7 mm	28.6 mm	42.3 mm

TABLE II - CORRELATION OF FEMORAL DIMENSION

Variables	Correlation coefficient	Statistical significance
A) Periosteal/Periosteal		
Femoral length vs. Head diameter	0.74	$p<0.0001$
Femoral head diameter vs. Extracortical	0.70	$p<0.0001$
Femoral length vs. Neck length	0.70	$p<0.0001$
Femoral head vs. Head position (C)	0.58	$p<0.0001$
Femoral head diameter vs. Neck-shaft angle	– 0.06	$p>0.06$
B) Endosteal/Endosteal		
Canal width (LT - 20 F) vs. Isthmus diameter	0.64	$p<0.0001$
Canal width (LT - 20 F) vs. Canal width (LT + 20 D)	0.65	$p<0.0001$
Canal width (LT + 20 D) vs. isthmus diameter	0.32	$p<0.0001$
Isthmus depth (K) vs. Femoral head position	0.14	$p<0.05$
C) Periosteal/Endosteal		
Canal width (LT - 20 F) vs. Femoral head diameter	0.58	$p<0.0001$
Isthmus diameter vs. Extracortical width (H)	0.49	$p<0.0001$
Isthmus depth (K) vs. Head offset	0.25	$p<0.005$
Canal width (LT - 20 F)	– 0.02	$p>0.05$

ful in the predictive sense.

The correlation coefficients (Tab. II) varied from –0.02 (canal width - LT -20,F) to 0.74 (femoral length vs. head diameter). Correlations were generally weaker between parameters characterizing the endosteal surface, suggesting a poor association between the internal and external profiles of th femoral canal; correlations were stronger betwee the parameters describing the width of the cana in the vicinity of the lesser trochanter.

Statistical analysis further indicated that the mediolateral width of the femoral canal at a point 2 mm distal to the lesser trochanter was the most predictable dimension of the femur and was estimated with an accuracy of 12%. This dimension was used as a basis for classifying specimens by canal size and estimating the inherent variability of all other dimensions (Tab. III). The accuracy of prediction of the dimensional parameters ranged from 12% to 32%. One of the most variable dimensions was the position of the isthmus within the medullary canal, which was only predictable to ± 33 mm (± 29 mm). The mediolateral width of the proximal canal and the medullary isthmus were predictable to within 17% (± 8.0 mm) and 27% (± 3.3 mm) respectively. This suggests that although significant correlations were present between the endosteal dimensions, the prediction of one dimension from another is impossible to the level of accuracy useful in implant design (1-2 mm).

Application of anatomic data to tha

The basic geometric data describing the proximal femur give guidelines for the functional shape of femoral components. This subject may be addressed in terms of restoration of the prosthetic hip (5) and factors affecting fixation of the femoral stem (6).

Restoration of the hip joint

To ensure the maintenance of leg length and to restore the original position for the balance of the abductor and joint forces, the original position of the femoral head center must be restored. For this purpose, the femoral component must be available in a range of neck lengths for each stem size. With cementless components it is impossible to adjust the leg length by changing the level of the femoral osteotomy, as in cemented fixation, and the stem position is determined by canal fit.

In 98% of the femora in this study, the femoral head was positioned from 38 to 67 mm (range 29 mm) superior to the level of the lesser trochanter, while the medial head offset varied from 30 to 57 mm (range 27 mm). Most of these differences were the results of changes in the neck-shaft angle and were not due to the inherent length of the femoral neck itself.

TABLE III - PREDICTION OF FEMORAL DIMENSIONS

Dimension	Accuracy of prediction 95% (confidence)	Percent of Average value
Head offset	±12.4 mm	±29%
Femoral head diameter	±7.6 mm	±17%
Femoral head position	±13.1 mm	±25%
Canal width (LT +20)	±8.0 mm	±17%
Canal width (LT)	±4.5 mm	±15%
Canal width (LT -20)	±2.5 mm	±12%
Isthmus width (mediolateral)	±3.3 mm	±26%
Extracorporeal width	±5.1 mm	±19%
Isthmus position	±31.0	±25%
AP canal width (osteotomy)	±5.3 mm	±23%
Femoral neck diameter	±5.4 mm	±32%
Isthmus width	±5.7 mm	±32%

To reproduce the original head center with a prosthesis, two alternate strategies may be considered. If the anatomic neck-shaft angle (range 117-141°) can be restored, a variation of only 11 mm (30 - 41 mm) in prosthetic neck length is necessary: alternatively, with a constant neck-shaft angle device that allows restoration of the leg length but not true head offset, a neck length over a range of 25 mm (25-50 mm) must still be available.

At present several hip replacement systems allow intraoperative alteration of the effective neck length by 10-14 mm; this adjustment does not allow restoration of the original head position in 15 - 30% of cases depending on the original stem configuration.

Femoral component fixation

It is extremely important to ensure a close match between the proximal cross-section of the bone and the femoral component (7, 8).

The size of the femoral cavity at the level of the femoral neck osteotomy is expressed in terms of its mediolateral and anteroposterior intracortical widths. At this level, the canal dimensions vary almost independently by 5 mm. This means that stem-bone contact within the metaphysis can only be obtained in discrete areas of the endosteal surface rather than over a substantial portion of the potential interface. For this reason, few femoral components truly fill the femoral canal in the lateral roentgenographic view.

In the lateral view, the femur presents an anterior bow in the diaphysis and a posterior bow in the metaphysis. This causes the femoral axis to pass approximately 8 mm posterior to the geometric center of the surface of the conventional femoral neck osteotomy (Fig. 3). As a result, femoral stems of a symmetrical cross-section are often implanted in close opposition to the posterior femoral cortex with a significant layer of interposed cancellous bone anteriorly. To provide a closer match to femoral anatomy, components may be designed with curved stems, with different implants for right and left femora.

The posterior curve of the proximal femur may be described as an arc that bisects the proximal femoral canal in the lateral view. A similar curve describing the anterior bow of the diaphysis joins the posterior arc close to the lesser trochanter. The angle of intersection of these anterior and posterior axes is an indicator of the lateral curvature of the proximal femur and ranges from 0° to 24° (average 10.7° ± 5.6°). Because of this variability, selection of a stem of fixed curvature does not ensure central placement of the implant within the metaphysis and may result in anterior impingement in some femora.

More femoral stems are designed to extend to the isthmus to stabilize component alignment and prevent varus migration. Consequently, selection of an optimal length for stems of different sizes depends on the relationship between the canal width and isthmus depth. This is of particular importance for all straight femoral stems which can lead to impingement between the implanted component and the cortical wall, as seen in lateral roentgenograms of components of excessive length (Fig. 4).

As the position of the femoral isthmus is variable, intramedullary reaming may be necessary to convert the canal into a parallel-sides section 70-80 mm long. The femoral stem can then be designed to engage that section in each femur. Measurements made this study show that for each canal size, a particular stem length exists that will allow the component to engage the isthmic region in virtually every bone.

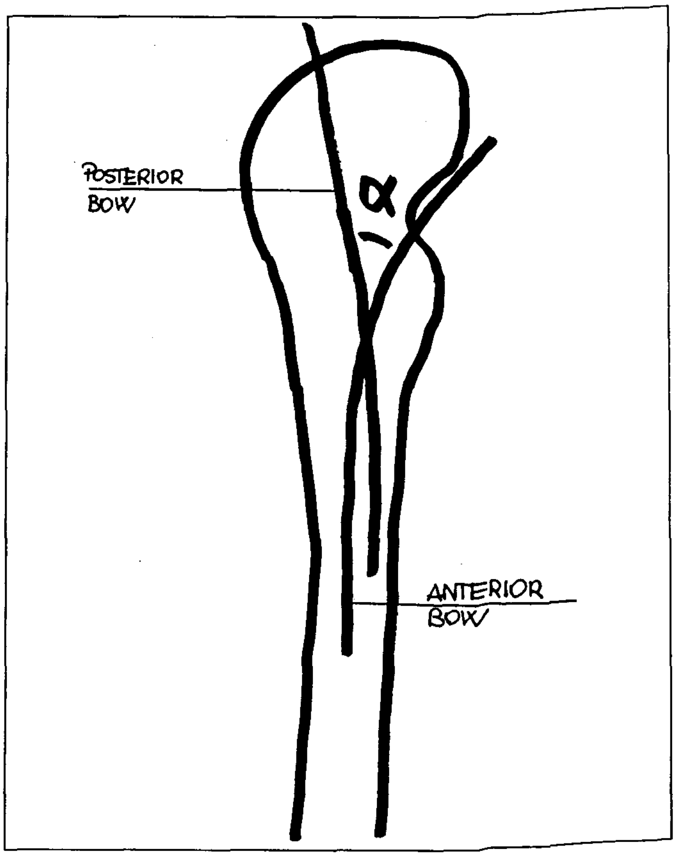


Fig. 3 - *Diagrammatic representation of the proximal femur showing the axes of the anterior and posterior bow.*

Patterns of the endosteal shape of the femur

As the joint reaction forces acting on the femoral head are orientated in the coronal plane, the relationship between the shape of the canal and its overall size is important for implant stability. The proportionality factors for the femoral population showed wide variations. Thus, on average, the silhouette of the femur roughly conforms to the proportions of those studied earlier (1). In contrast, the femoral canal was distinctly non-proportional. Over the anatomic range the enlargement of the endosteal dimensions ranged from 20% for the depth of the canal isthmus to 79% for the width of the medullary canal 20 mm distal to the lesser trochanter.

Over the same anatomic range, the canal flare index was broadly distributed from 2.4 to 7.0, average 3.8 ± 0.74. The canal flare index showed a

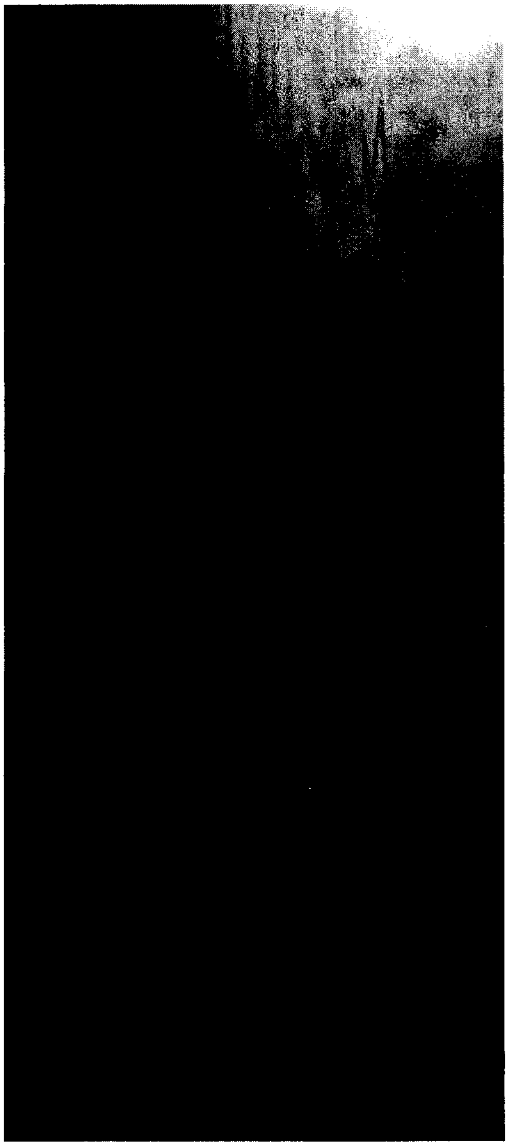


Fig. 4 - *In lateral roentgenograms excessive length of components.*

skewed normal distribution with maximum incidence at 3.4; however, a substantial number of bones had indices between 2.8 and 4.8. Comparison of subjective descriptions of canal shape with actual values of the index showed that canal flare indices of less than 3.0 described stovepipe canals, 3.0-4.7 normal canals, and 4.7-6.5 canals with a champagne-fluted appearance.

Canal flare indices were also correlated with canal size, expressed as the mediolateral width 20 mm distal to the lesser trochanter. A weak inverse correlation between these dimensions indicated a general tendency for bones with small canals to have a champagne-fluted shape and those with the largest canals to be of the stovepipe configuration. A strong inverse relationship was noted between the canal flare index and size of the femoral isthmus that indicated that most femora with a wide isthmus have a correspondingly straighter canal profile and are not simply larger bones of one basic endosteal geometry.

Age-related changes in canal shape

Examining the canal flare index by bone age, we observed a significant shift in the canal shape, ranging from a pronounced proximal canal flare in young bones to straighter canals in older femora. Almost every bone younger than 55 years had an index of 4.0 or less. This geometric change is related to a widening of the cortex with aging, which averaged 1.3 mm per decade at the isthmus. This was associated with a compensatory thinning of the cortex that became apparent in bones older than 60 years.

Development of femoral stems matches endosteal geometry

Given the variability of the size and shape of the proximal femur, methods of component fixation must be able to accommodate variations in bone implant fit regardless of whether fixation is through tissue ingrowth, press fitting, or the use of bone cement.

As it is impossible to predict femoral geometry (line to line or surface to surface), contact between the implant and bone is unexpected except in discrete areas of interface.

In any implant system, a direct relationship exist between the fitting accuracy of each femoral canal and the number of different component geometries available to the surgeon. The anatomic range of femoral geometry may be represented by a finite set of unique canal shapes called morphotypes. Each morphotype was selected to describe the medial and lateral endosteal contours of each proximal femur to an accurancy of 1 mm. Based on the observed geometry of bones of different sizes, some

45 morphotypes were initially derived. Only 17 of these occurred with an incidence of more than 1% in the study population. Thus if proximal fitting of implants to the bone were necessary to within 1 mm, a system of 17 separate components would be necessary to match the anatomic form of the femora (Fig. 5).

DISCUSSION

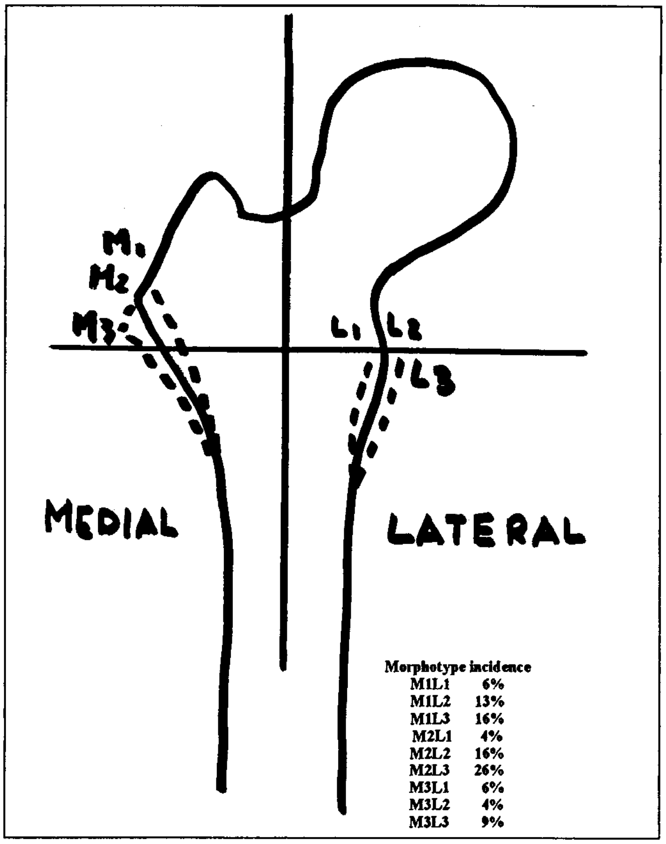
Many studies (9, 10) have demonstrated that direct support of the femoral component by the strongest available bone is possible only if instruments and implants are available that closely approximate the endosteal geometry of the surface.

Long-term follow-up reviews of cemented THA show a strong correlation between inadequate resection of cancellous bone and aseptic loosening of the femoral component; this is not surprising because the load bearing capacity of the cement-bone interface is determined by the strength of cancellous bone (11, 12).

In cementless implants, a universal conclusion is that immediate stable fixation is fundamental to the successful hip arthroplasty. Moreover, the degree of implant-bone fit appears to correlate with the quality of the clinical outcome in cementless procedures. Patients with close-fitting femoral components reported higher overall scores on the d'Aubigné scale than those with less adequate bone-implant matches (13). Studies comparing three designs of cementless implants point out that subsidance of up 10 mm, occurring in 1/4 of all components, was directly correlated to a mismatch of implant and canal sizes (14). Another observation in the follow-up of the series of cementless components is the improvement of clinical results with the availability of the greater range of sizes of femoral components to match endosteal geometry (15, 16).

Analysis of femoral anatomy shows that femoral components must accommodate anatomic variability to restore the normal biomechanics of the hip joint. This requirement stands in marked contrast to the relatively limited capacity of modern design of THA.

The variability of endosteal morphology is consistent with the assumption that femoral geometry is determined by a large number of genetic and environmental factors, including age, sex, race and lifestyle. These influences lead to unique endosteal geometries as characteristic of each individual as any other feature of human anatomy.



Morphotype	incidence
M1L1	6%
M1L2	13%
M1L3	16%
M2L1	4%
M2L2	16%
M2L3	26%
M3L1	6%
M3L2	4%
M3L3	9%

Fig. 5 - *Diagrammatic representation of one method of derivation of canal morphotype. Bones of a given canal diameter are classified according to their medial and lateral endosteal profiles.*

This study confirms that there is no proportional relationship between the size and shape of the medullary canal. Thus implants cannot be designed on the basis of an average canal size scaled for larger and smaller canals in the anatomic range.

The distribution of the canal flare index shows that characterization of the femur by a single canal shape is impossible and the correlation between the distal and proximal dimensions of the medullary canal is poor. This means that if metaphyseal load transfer is to be optimized, femoral components must be selected on the basis of proximal canal fit and not the diamete: of the isthmus.

Many studies attest to the effect of aging on femoral morphology. Endosteal changes occur predominantly in the diaphysis in which cortical thinning leads to medullary expansion from about the fourth decade of life onward. This has several implications for THA. The stabilization of the femoral component relies on a balance of proximal and distal load transfer from the implant to femur. This relative support depends on many factors including the prosthesis fit to the bone and the rigidity of the zones of fixation between the stem and the femur. Close stem fit reduces the concentration of proximal stresses at the implant-bone interface. If, however, medullary expansion occurs with aging, the resistance of the prosthesis to medial migration can be undermined, resulting in an increased rate of mechanical loosening and subsidance.

A practical consequence of the effect of age on canal geometry is that femoral components of different overall shapes may be needed for patients of different ages. As the average flare index decreases from 4.6 to 3.4 for patients from 50 to 75 years, femoral stems designed for cementless fixation must be wider proximally for a given distal diameter.

This suggests that no one basic design of a femoral stem is compatible with optimal performance of both cemented and cementless THA.

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