The Effects of Total Hip Arthroplasty on the Structural and Biomechanical Properties of Adult Bone

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KEY WORDS bone; mechanical loading; senescence; periosteal apposition

ABSTRACT The responsiveness of bone to mechanical stimuli changes throughout life, with adaptive potential generally declining after skeletal maturity is reached. This has led some to question the importance of bone functional adaptation in the determination of the structural and material properties of the adult skeleton. A better understanding of age-specific differences in bone response to mechanical loads is essential to interpretations of long bone adaptation. The purpose of this study is to examine how the altered mechanical loading environment and cortical bone loss associated with total hip arthroplasty affects the structural and biomechanical properties of adult bone at the mid-shaft femur. Femoral cross sections from seven individuals who had undergone unilateral total hip arthroplasty were analyzed, with intact, contralateral femora serving as an approximate internal control. A comparative sample of individuals without hip prostheses was also included in the analysis. Results showed a decrease in cortical area in femora with prostheses, primarily through bone loss at the endosteal envelope; however, an increase in total cross-sectional area and maintenance of the parameters of bone strength, I_x, I_y, and J, were observed. No detectable differences were found between femora of individuals without prostheses. We interpret these findings as an adaptive response to increased strains caused by loading a bone previously diminished in mass due to insertion of femoral prosthesis. These results suggest that bone accrued through periosteal apposition may serve as an important means by which adult bone can functional adapt to changes in mechanical loading despite limitations associated with senescence. Am J Phys Anthropol 138:221–230, 2009. ©2008 Wiley-Liss, Inc.

The relationship between mechanical loading and bone mass is well substantiated. The application of beam theory to the study of mechanical properties of human long bones has proven to be informative in both clinical (Beck et al., 1996; Ruff, 2003; Taaffe et al., 2003; Janz et al., 2004) and archaeological (Ruff and Hayes, 1983a,b; Ruff, 1987; Lazenby, 1998; Larsen et al., 2001; Stock and Pfeiffer, 2001; Holt, 2003; Rhodes and Knusel, 2005) contexts. Of particular interest is the age-specificity of bone response to changes in mechanical loading. Age-associated decline in osteoblastic function and cellular responsiveness to hormonal and other chemical signals substantially hinder bone's ability to respond to mechanically induced strain (Pearson and Lieberman, 2004); consequently, physical activity during childhood and adolescence has a greater influence on bone mechanical properties than physical activity during later life (Turner and Robling, 2003). A better understanding of adult bone adaptation is important for several reasons, including identification of metabolic bone disease and fracture risk, and explication of mechanisms affecting interpretations of long bone adaptations in both clinical and archaeological contexts.

Total hip arthroplasty presents a unique opportunity to observe adult bone adaptation, as the procedure engenders changes in the mechanical loading environment to which bone responds. Several studies have assessed variables that relate to changes in the bone density and structure of skeletally mature individuals following prosthetic implantation. There is general agreement that bone mass decreases at the periprosthetic region postoperatively, with peak bone loss occurring at the proximal femur and declining distally. Periprosthetic bone loss averaging 5–45% and ranging as high as 52% has been shown in retrospective, prospective, and postmortem studies (Sumner and Galante, 1992; Toni et al., 1992; Maloney et al., 1996; Engh et al., 1999; Yamaguchi et al., 2000; Pitto et al., 2001; Sychterz et al., 2001; Tanzer et al., 2001; Wright et al., 2001; Aldinger et al., 2003; Schmidt et al., 2003, 2004). Decreases in bone mass have also been demonstrated for other skeletal sites outside the periprosthetic region. Peripheral and axial bone loss ranging from 3 to 16% after total hip arthroplasty have been reported for the middle and distal femur (Lindberg and Nilsson, 1984; Torchia and Ruff, 1990; Adolphson et al., 1993; Bryan et al., 1996), tibia (Ruegsegger et al., 1986), calcaneus (Hirano et al., 2001), and lumbar vertebrae (Black et al., 1985; Souen et al., 1992; Adolphson et al., 1994; Hirano et al., 2001).

Several factors may influence the extent of bone loss after total hip arthroplasty, including sex, weight, mechanical loading, disease, preoperative bone state, medication, duration of implantation, and the design criteria of the implant, such as prosthesis size, shape, coating, and fixative. Of these, it is widely accepted that mechanical loading is the most important (Bobyn et al., 1990; Engh et al., 1992; Weinans et al., 1994; Huiskes, 1995; Kerner et al., 1999; Lim et al., 1999). Insertion of a femoral prosthesis, regardless of its specific characteristics,

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has been shown to result in marked alterations in the mechanical loading environment. Postoperatively, ground reaction and muscle forces are no longer solely transferred to bone at the periprosthetic level, but rather are shared by the bone and the prosthesis, resulting in a phenomenon known as "stress shielding" (Bobyn et al., 1992; Dujovne et al., 1993; Skinner et al., 1994; Sychterz et al., 2001). Studies directly measuring *in vivo* strain demonstrate that this alteration results in significantly lower levels of strain in the region (Cheal et al., 1992; Decking et al., 2006). Periprosthetic bone loss observed after total hip arthroplasty is primarily attributable to this phenomenon.

Stress shielding alone, however, does not account for bone loss observed in other regions of the body; pre and postoperative gait alteration and surgery induced trauma have also been shown to result in substantial changes in the mechanical loading environment. Prior to implantation of a femoral prosthesis, functional adaptations in gait due to pain, weakness, or instability resulting from an abnormal hip joint may reduce the amount of stress in affected femora (Kiratli et al., 1991; Adolphson et al., 1993; Kale et al., 1995; Sauer et al., 1996; Lim et al., 1999). Because femoral loading is defined primarily by muscle and joint contact (Duda et al., 1998), the location and extent of surgical trauma to muscle can alter the mechanical forces to which the femur is exposed (Zimmerman et al., 2002; Heller et al., 2003; Bitsakos et al., 2005; Perka et al., 2005). Postoperatively, reduction in skeletal loading results in marked declines in intersegmental and muscle forces, particularly in the implanted limb (Adolphson et al., 1993; Bryan et al., 1996). These alterations significantly contribute to periprosthetic, axial, and peripheral bone loss. Although there is ample evidence to support the relationship between changes in mechanical loading and bone loss after total hip arthroplasty, little is known about the specific details of this interaction, because most research addresses the clinical outcome of implantation rather than questions regarding general bone functional adaptation.

This study presents the results of an analysis of cross-sectional geometric properties in a sample of individuals who had undergone unilateral total hip arthroplasty. The method employed uses human cadaver samples for which both implanted and intact, contralateral femora were available for each individual. The study has two primary purposes: first, to make specific quantifications of cortical bone changes in the mid-diaphyseal femur after prosthetic implantation; and second, because bone response after implantation is primarily a mechanically mediated phenomenon, to test the hypothesis that adult bone will respond by adapting to its loading environment, despite decline in adaptive potential. Specifically, this study examines how the altered mechanical loading environment and cortical bone loss associated with total hip arthroplasty affects the geometric properties of bone at the mid-shaft femur in skeletally mature individuals.

MATERIALS AND METHODS

The study sample was drawn from femora harvested from decedents at the George Washington University Medical School Department of Anatomy between 1972 and 1982. A total of 14 paired femora from seven individuals who had undergone unilateral total hip arthroplasty with cemented Charnley prostheses were analyzed. Implanted prostheses extended into the proximal one-third of femora; however, the exact location of the stem tip in each individual is unknown. The sample consisted of one male and six females whose ages ranged from 69 to 95 years at death (mean, 80 years). In each case, the hip replacements were clinically successful at the time of death.

TABLE 1. Age and sex of comparative and prosthesis samples

Specimen	Age/Side[a]	Sex
Prosthesis-1	82/R	Female
Prosthesis-2	70/L	Female
Prosthesis-3	79/R	Female
Prosthesis-4	95/R	Female
Prosthesis-5	84/R	Female
Prosthesis-6	69/R	Male
Prosthesis-7	84/R	Female
Comparative-1	60/–	Male
Comparative-2	39/–	Female
Comparative-3	52/–	Male
Comparative-4	82/–	Female
Comparative-5	54/–	Female

[a] Side implanted with femoral prosthesis.

A comparative sample of 10 paired femora from five individuals without hip prostheses was also included in the analysis. Individuals comprising the comparative sample were drawn from the same collection as those with prostheses. Although little bilateral asymmetry exists in the lower limb (Auerbach and Ruff, 2006), comparing intact, paired femora from individuals of the same collection allows for the normal relationship between paired femora to be determined. The sample consisted of two males and three females whose ages ranged from 39 to 82 years at death (mean, 57 years). In both prosthetic and comparative samples, all individuals died of natural or traumatic causes and no evidence of metabolic abnormalities were present. The specimens used in both samples are listed in Table 1.

Direct cross-sectioning of cortical bone was used to quantify the area and distribution of bone. Because invasive sectioning of bone serves as a direct measure of bone structure, it is both highly accurate and reliable. Following this method, a total of 24 transverse cross sections of bone from mid-shaft femora were obtained (see Fig. 1). Intact, contralateral femora served as an approximate internal control; because comparisons were made between femora of the same individuals, the use of the contralateral limb advantageously controls for potential factors that may influence bone dynamics (Maloney et al., 1996; Engh et al., 1999). From each individual, the mid-shaft of the left and right femur was sectioned ($\sim 80\ \mu$m) on a Buehler® Isomet low-speed petrographic diamond-bladed saw and mounted on microscope slides. Complete transverse cross sections were scanned using a flatbed scanner and then imported into image editing software programs. Scanned cross-sectional photos were transformed into flattened, black and white, bitmap images using Adobe® Photoshop® Elements. Reference axes of specimens were predefined (i.e., AP, ML), allowing second moments of area about anatomical axes (I_x, I_y) to be accurately measured. The resulting sections were used to calculate standard cross-sectional properties using a PC version of SLICE (Nagurka and Hayes, 1980) loaded into Scion Image (Release Beta 4.0.2). To provide further information on bone structural properties, femoral sections derived from the prosthetic sample

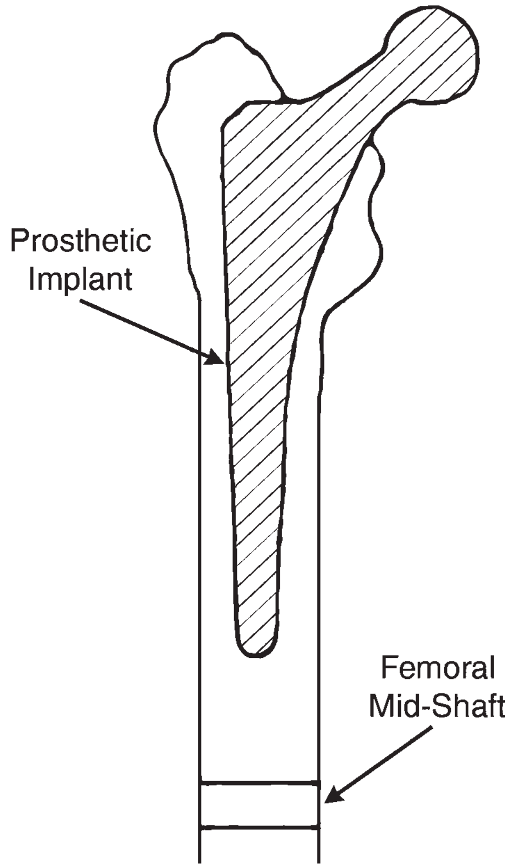

Fig. 1. Femoral region examined (figure indicates relationship of sample site to prosthesis in implanted limbs).

were read on a compound microscope at 10× and 20× magnification, allowing bone histomorphology to be assessed.

Cross-sectional properties of the mid-shaft femur were used to assess adult bone adaptation following total hip arthroplasty. The application of certain engineering principles in biological contexts is particularly suited to biomechanical analyses of long bone cross-sectional geometry (Ruff and Hayes, 1983a). This is because a long bone may be considered a hollow beam, thus standard beam theory can be used to predict rigidity and strength under loading. Geometric properties are determined from the amount and distribution of bone in a section. Although variables derived from a purely geometric analysis do not take into account possible differences in bone material properties, such differences are likely to be relatively small (Erickson et al., 2002); therefore, geometric analysis presents a sound method by which resistance to mechanical loads may be evaluated (Ruff, 2008). The standardization of cross-sectional properties to body size is essential in studies comparing skeletal elements among individuals (Ruff et al., 1993). The current study, however, compared femora from the same individual, so scaling of cross-sectional properties was not necessary.

The cross-sectional properties calculated include total sub-periosteal area (TA) and cortical area (CA). Medullary area (MA) was then determined by subtracting CA from TA. Cross-sectional area not only serves as an accurate measure of bone structure, but also as an appropriate dimension for evaluating internal resistance to tensile and compressive loads (axial loading). Long bones, however, are subjected to significant bending as well as pure axial loading because of bone curvature and muscle action (Ruff and Hayes, 1983a; Currey, 1984). The second moment of area (I) is the most important cross-sectional characteristic for evaluating bone strength under bending. Therefore, second moments of area I_x and I_y, proportional to bending rigidity in the anterior–posterior and medial–lateral planes, respectively, were used to evaluate resistance to bending. The polar second moment of area, J, was also calculated. In typical cross-sections, J has been shown to provide the most accurate estimate of average bending and torsional rigidity (Lieberman et al., 2004).

Statistical analysis was conducted using the software package SPSS. A paired t-test was used to test for differences in cross-sectional properties (TA, CA, MA, I_x, I_y, J) between contralateral femora in both prosthetic and comparative samples.

RESULTS

Results of the geometric analyses are given in Table 2 and Figure 2. Compared with their intact, contralateral sides, implanted femora have significantly lower cortical

TABLE 2. Paired-samples t*-test for prosthetic and comparative samples (left–right)*

		Paired samples tests							
		Paired differences							
					95% Confidence interval of the difference				
		Mean	Std. deviation	Std. error mean	Lower	Upper	t	df	Sig. (2-tailed)
Pros	Pros CA − Intact CA	−0.33286	0.24777	0.09365	−0.56201	−0.10371	−3.554	6	0.012
Pros	Pros MA − Intact MA	−0.63000	0.35865	0.13556	0.29830	0.96170	4.647	6	0.004
Pros	Pros TA − Intact TA	−0.29000	0.20494	0.07746	0.10046	0.47954	3.744	6	0.010
Pros	Pros I_x − Intact I_x	−0.04571	0.05255	0.01986	−0.00289	0.09432	2.301	6	0.061
Pros	Pros I_y − Intact I_y	−0.01143	0.05928	0.02241	−0.04340	0.06625	0.510	6	0.628
Pros	Pros J − Intact J	0.05714	0.06849	0.02589	−0.00620	0.12048	2.208	6	0.069
Comp	Intact CA − Intact CA	−0.03400	0.05177	0.02315	−0.09828	0.03028	−1.469	4	0.216
Comp	Intact MA − Intact MA	0.08000	0.08426	0.03768	−0.02462	0.18462	2.123	4	0.101
Comp	Intact TA − Intact TA	0.04600	0.04930	0.02205	−0.01521	0.10721	2.087	4	0.105
Comp	Intact I_x − Intact I_x	−0.00800	0.09985	0.04465	−0.13198	0.11598	−.179	4	0.867
Comp	Intact I_y − Intact I_y	0.05000	0.08660	0.03873	−0.05753	0.15753	1.291	4	0.266
Comp	Intact J − Intact J	0.04200	0.11032	0.04934	−0.09498	0.17898	0.851	4	0.443

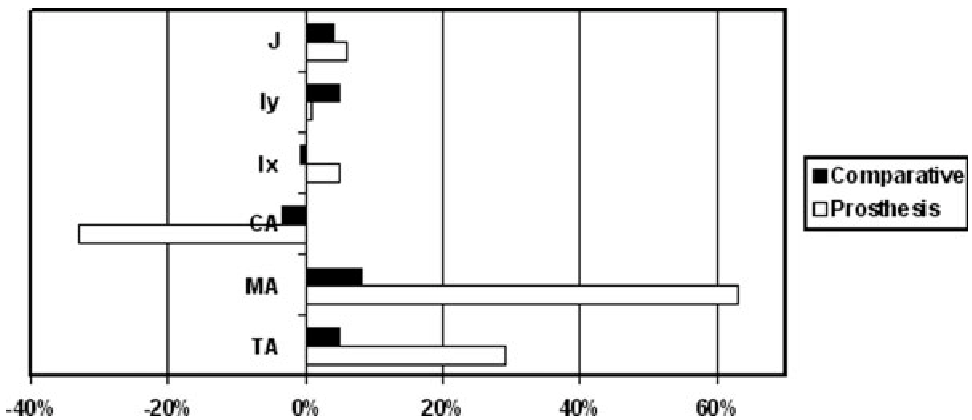

Fig. 2. Mean percent differences, implanted-contralateral of prosthesis sample, and left–right femora of comparative sample.

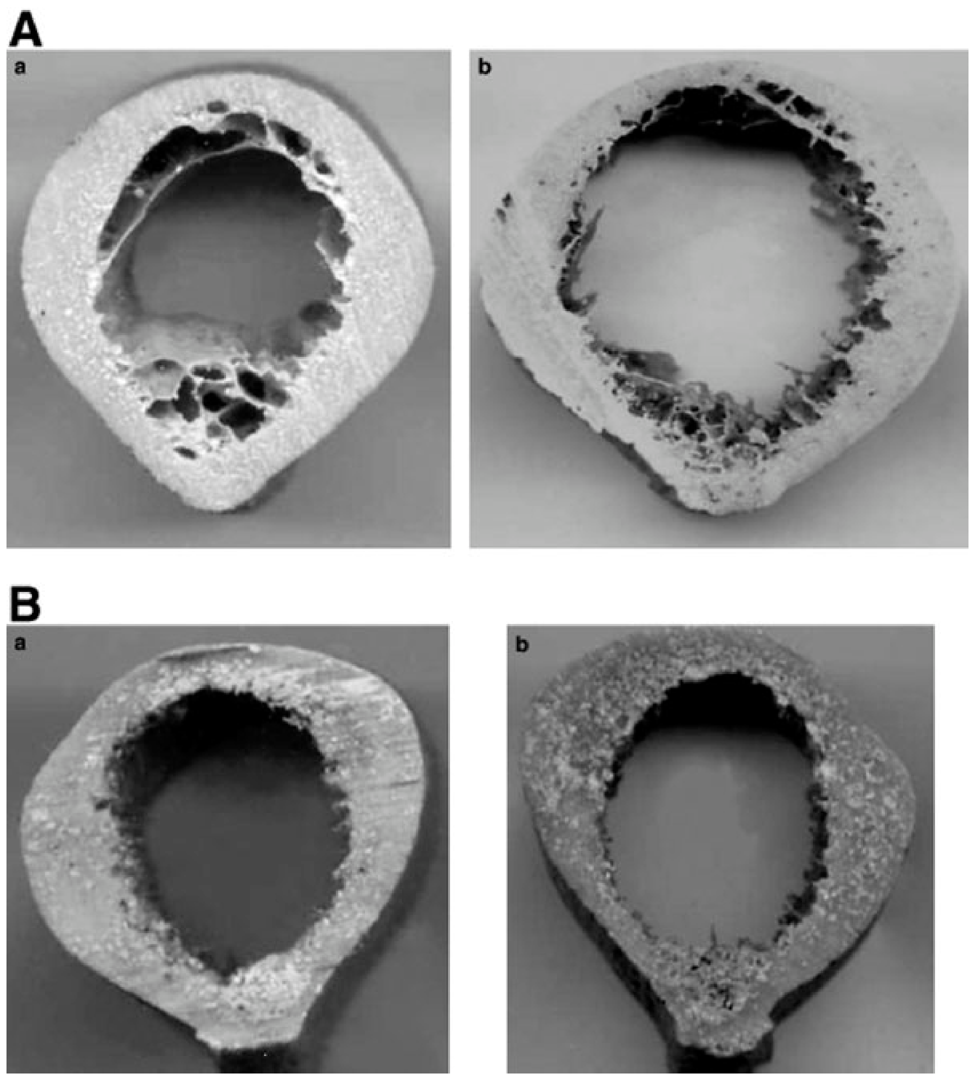

Fig. 3. Comparisons of right and left femora demonstrating the significant difference in cross-sectional properties observed among individuals implanted with femoral prostheses compared to those in the comparative sample. (**A**) Female aged 79 years at death (Pros-3); (a) intact and (b) prosthesis. (**B**) Female aged 82 years at death (Comp-4); (a) intact and (b) intact.

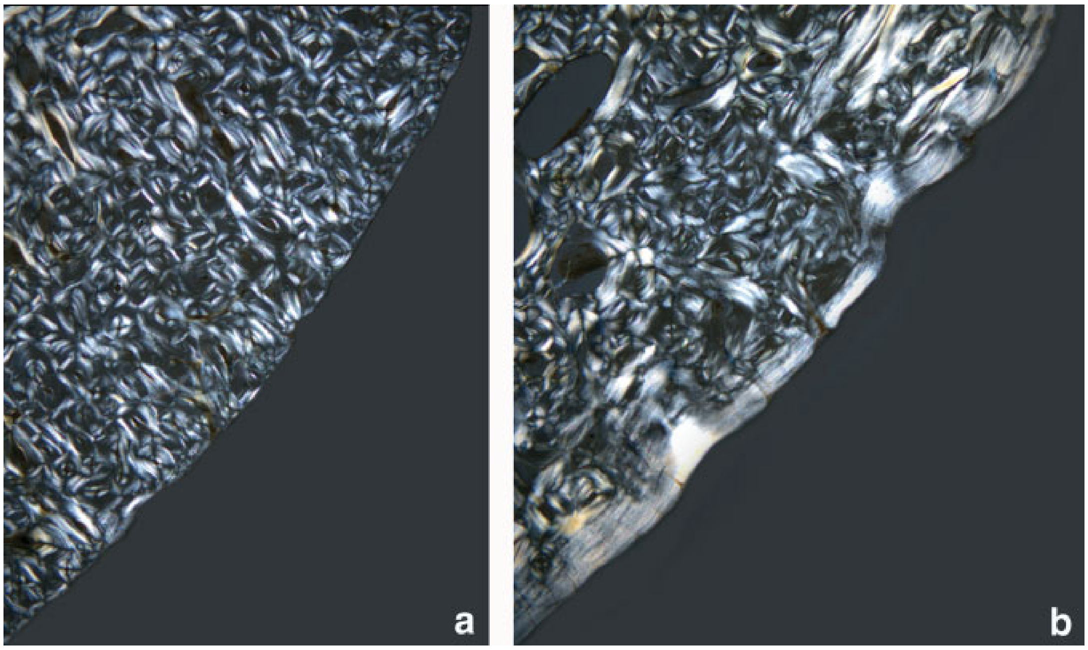

Fig. 4. Polarized histological structure of the femoral posterolateral quadrant. Female aged 79 years at death (Pros-3). (a) intact and (b) prosthesis. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

area (CA), whereas marrow area (MA) and total area (TA) are significantly higher (see Fig. 3). The percentage increase in MA is larger than that in TA, thus explaining the net loss of cortical bone in femora with implanted prostheses. Although not statistically significant, both I_x ($P = 0.061$) and J ($P = 0.069$) show a slight increase in implanted femora, whereas no difference in I_y is observed. Because of the increase in TA in implanted femora, cortical bone is placed further from the neutral axis, resulting in a cross section that is more resistant to bending and torsional stresses. This illustrates that bone strength is similar between implanted and contralateral femora despite the decline in CA observed in the former.

All implanted femora showed signs of increased periosteal activity compared with their intact contralateral sides when examined at the histological level (see Fig. 4). This was manifest as additional periosteal lamellae in implanted femora. Periosteal apposition was found to have varied in magnitude around the circumference of the bone, with the greatest activity found at the medial and lateral aspects of femora. Periosteal activity of intact contralateral femora was typical of normal age-associated periosteal apposition.

In contrast to the prosthetic sample, no significant change is present in any cross-sectional geometric property between left and right femora of the comparative sample (see Fig. 2). In other words, no trend was observed, indicating that one side was consistently favored over the other. Therefore, similar to previous studies, little bilateral asymmetry is present in the lower limb. This lack of bilateral asymmetry in normal, intact femora from the same collection highlights the significance of differences found between implanted and contralateral, intact femora.

A substantial amount of variation in changes in cross-sectional properties is observed when individuals of the prosthetic sample are compared separately. For example, in individual two, a 70-year-old woman, there was a 21% decrease in cortical bone area. In contrast, in individual seven, an 84-year-old woman, only a 4% decrease in cortical bone area was observed. This same trend is reflected at the microscopic level, where the amount of periosteal apposition and extent of remodeling in periosteal lamellae was also found to vary. The observed range of variation among individuals is most likely due to differences in time since implantation and extent of postoperative physical activity. Because of the lack of information on each individual, the focus of this study is on the group as a whole, so individual variation will not be discussed any further.

DISCUSSION

Previous studies have demonstrated a significant reduction in the adaptive potential of bone after skeletal maturity is reached. Age-associated declines in osteoprogenitor cells and cellular responsiveness to hormones, mechanical stimuli, and biochemical nutrients, coupled with age-associated degenerative loss of skeletal muscle mass render bone less sensitive to physical activity (Bertram and Swartz, 1991; Tanaka and Liang, 1996; Nishida et al., 1999; Frost, 2000; Stanford et al., 2000; Pearson and Lieberman, 2004). This has led some to question the importance of functional adaptation in adult bone (Lovejoy et al., 2003). The results of this study, however, suggest that aging adults still retain the ability to structurally adapt to alterations in mechanical loads. This is manifest in two primary alterations in the structure of implanted femora: a decrease in area and the redistribution of cortical bone further from the neutral axis.

Numerous studies have demonstrated adaptive changes in bone following total hip arthroplasty. The majority of these reports have shown that implantation of a femoral prosthesis is associated with decreased stress and subsequent bone loss. Although clinical concern with implant efficacy has resulted in the periprosthetic region being the primary site assessed by most researchers, bone loss has also been reported among peripheral and axial weight-bearing sites. Adolphson et al. (1993), for example, reported a 9% loss of muscle volume and cortical bone at the mid-shaft of implanted femora 11 years after surgery. Similarly, Bryan et al. (1996) found patients with an excellent clinical result 10 years after unilateral total hip arthroplasty had 16% less bone in their affected limbs distal to the prosthesis compared with their contralateral limbs. These findings are in agreement with the results of this study, specifically that the area of bone within mid-shaft cross sections was significantly lower in femora subjected to prosthetic implantation.

Several factors may have contributed to the observed reduction in cortical area, including temporary immobilization during recovery, musculoskeletal trauma, and asymmetries in loading due to pre and postoperative gait alteration (Bryan et al., 1996; Lim et al., 1999; Perka et al., 2005). A marked reduction in stress and strain would have resulted from such alterations to the mechanical loading environment. Because loads on bones cause strains that generate signals that some cells can detect and to which they or other cells respond, this reduction can lead to a cascade of events. According to Frost's (1987) mechanostat model, genetically determined threshold ranges of these signals help to control modeling and remodeling. If peak strain magnitudes do not surpass a minimum effective strain required to suppress remodeling (MES_r), the remodeling rate is accelerated. Empirical evidence at the biochemical level for increased remodeling following total hip arthroplasty has been demonstrated by Whitson et al. (2002) who, through the measurement of indicators of bone formation and resorption, found an uncoupling of bone turnover following hip replacement surgery. In this study, it is likely that strains dropped below the MES_r in implanted femora, leading to accelerated remodeling. In accordance with disuse-associated bone loss, this appears to have occurred at the endosteal envelope, resulting in a net loss of cortical bone.

In implanted femora, endocortical expansion alone would reduce whole-bone strength. Bone strength, however, depends not only on the addition or subtraction of bone, but also its structural characteristics. In this study, although total area expansion was found to only be a fraction of that observed in marrow area, a larger bone circumference requires less bone area to maintain mechanical competence (see Fig. 5), as the bending strength of a particular area of bone is proportional to the fourth power of its distance from the neutral axis (Davison et al., 2006). This may explain the observed lack of difference in the measured parameters of bending rigidity (I_x, I_y, J) between intact and implanted femora. There are several processes that may account for the increase in total area that we have demonstrated. First, the period of decreased strain and bone loss associated with total hip arthroplasty may have been followed by a period of increased strain. After hip replacement, recovery is achieved in about 3–6 months (Engh, 2003). This is accompanied by pain relief and some functional restoration in most patients (Lim et al., 1999). Consequently, many studies have reported a general increase in bone mass after a period of initial bone loss at peripheral and axial sites, which appears to be dependent upon degree of postoperative physical activity (Lindberg and Nilsson, 1984; Ruegsegger et al., 1986; Adolphson et al., 1994; Hirano et al., 2001). Second, implantation of a femoral prosthesis may result in postoperative redistribution of strain levels that produce stress shielding of the proximal femur, with a high proportion of the shielded stress being transmitted from the implant to bone distally (Svesnsson et al., 1977; Jacob and Huggler, 1980, McBeath et al., 1980; Weightman et al., 1987). This has the effect of reversing the pattern of strain found in an intact femur, in that maximum strain occurs around and distal to the tip of the prosthesis rather than proximally when loaded (Oh and Harris, 1978). Research has shown this result in proximal resorption and distal cortical hypertrophy in many cases (Ritter and Fechtman, 1988; Cohen and Rushton, 1995; Reitman et al., 2003).

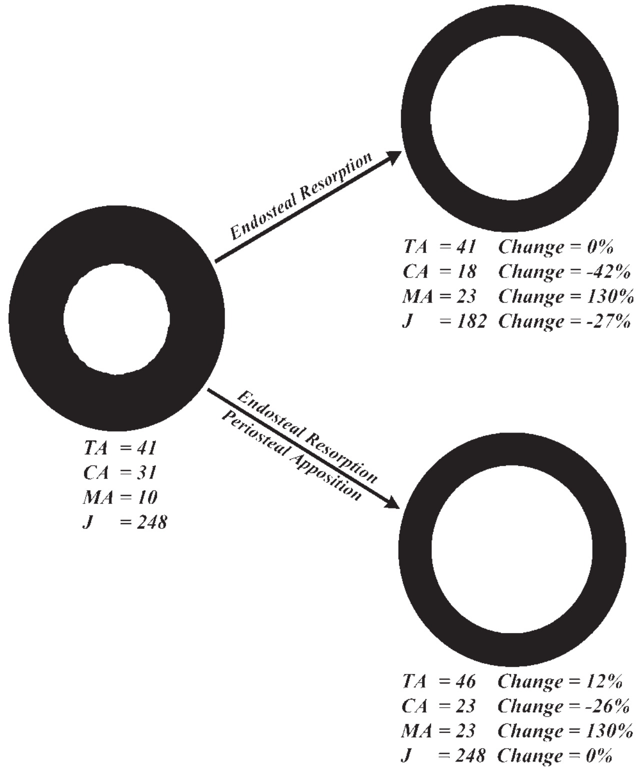

Fig. 5. Redistribution of cortical bone further from the neutral axis contributes disproportionately to bone strength: cross-section before changes (left); Alt 1, endosteal resorption leads to a decline in bone strength (top); Alt 2, minimal periosteal apposition compensates for the decline in bone strength caused by endosteal resorption (bottom).

It is therefore conceivable that a combination of increased physical activity and postoperative strain redistribution resulted in larger bone strains at the mid-shaft femur. This, in conjunction with previous endocortical remodeling, would have resulted in bone deformation exceeding an acceptable level, therefore challenging bone stability. As an organ, however, bone is primarily concerned with maintaining its biomechanical integrity (Frost, 2000). The function of cells responsible for mechanically adaptive modeling and remodeling is to ensure that bone structural and material properties are appropriate to the applied load. Despite age-related decline in osteogenic potential, this appears to have been partly accomplished by periosteal expansion of the cortical shell in this study, resulting in additional periosteal lamellae primarily along the mediolateral plane. There are several reasons for this pattern of differential periosteal apposition. First, the placement of the hip joint in relation to the body center of gravity in the female component of the sample may have contributed to greater mediolateral loading of the lower limb (Ruff and Hayes, 1983b). Second, the predilection for walking rather than running and lower levels of physical activity found generally among older individuals (Sinclair and Dangerfield, 1998) may have lead to more mediolaterally distributed cross sections (Lovejoy et al., 1976).

Similar findings regarding specific structural changes are uncommon in the clinical literature on total hip arthroplasty. This may be partly due to the predominant use of noninvasive methods, which can lack detail and precision. This point is demonstrated by the use of dual energy X-ray absorptiometry (DEXA). Using this technique, diminished bone mineral content following total hip arthroplasty could be explained by a decrease in cortical bone area, an increase in cortical porosity, or both. Because DEXA fails to distinguish between bone structural and material properties, it does not permit the full determination of the processes that account for postoperative change. When more accurate methods have been used, contrasting results have been reported. Torchia and Ruff (1990), for example, through the direct cross-sectioning of cortical bone, found a general decrease in endosteal and total area at the middle and distal diaphysis of implanted femora, while cortical area and average bending rigidity increased. It is important to note, however, that three of the four individuals included in the analysis were implanted with uncemented prostheses, which can have a significant effect on postoperative stress transfer due to greater stem mobility. When considered alone, the individual with the cemented prosthesis clearly departed from the pattern observed in the uncemented sample, demonstrating a marked decline in I_y around and just distal to the prosthesis stem tip. Therefore, differences in lifestyle and design criteria may partially explain the discrepancy found with the results of this study.

These findings support the idea, first suggested by Smith and Walker (1964), that increases in periosteal area and geometric properties are adaptations to mechanically induced strain. Although Smith and Walker's findings have been open to some debate (Trotter et al., 1968), many studies have since produced supporting evidence of the crucial role of periosteal apposition in the maintenance of biomechanical integrity in the aging skeleton (Epker and Frost, 1966; Ruff and Hayes, 1983a; Beck et al., 1992; Bouxsein et al., 1994; Seeman, 2002; Ahlborg et al., 2003; Orwoll, 2003). One way periosteal apposition may be stimulated is through the relatively greater strain experienced at the bone periphery when loaded due to bone loss at the endosteal envelope (Lazenby, 1990; Russo et al., 2003; Seeman, 2003). This has led to the general acceptance of the hypothesis that increased bone mass accumulated through periosteal apposition provides mechanical compensation for the reduction in mass resulting from endosteal resorption (Lazenby, 1990). In this study, a similar process is thought to have occurred due to surgical implantation of a femoral prosthesis. Namely, periosteal apposition appears to be an adaptive response to increased strains caused by loading a bone diminished in mass by endocortical resorption.

The results of this study have implications for bone's ability to adapt after skeletal maturity is reached. The responsiveness of modeling and remodeling to loading is greatly decreased in adult bone. The results of this study, however, seem to indicate that bone may have the capability to alter its structure to an appreciable degree, even after skeletal maturity is reached. This is in accordance with previous research that has demonstrated the retention of at least some osteogenic potential in most aging adults (Kannus et al., 1995; Kerr et al., 1996; Kontulaenen et al., 2002; Valdimarsson et al., 2005). Periosteal apposition appears to play a central role in this process. Because the redistribution of cortical bone further from the neutral axis contributes disproportionately to bone rigidity, small increases in circumference significantly contribute to bone strength. Therefore, periosteal apposition provides an important means by which bone may adapt to functional alterations in mechanical loading despite limitations associated with senescence.

CONCLUSIONS

In this study, analysis of 14 implanted and contralateral mid-shaft femora presents clear evidence of bone functional adaptation in skeletally mature individuals following total hip arthroplasty. The results corroborate previous studies that have elucidated some of the factors that govern the incidence and severity of bone resorption after hip replacement, both in clinical and laboratory settings. Less appreciated, however, is the increase in periosteal area and subsequent maintenance of bone strength observed. These changes in femoral cross-sectional geometry have implications for the responsiveness of bone to mechanical stimuli after skeletal maturity is reached, as well as for the public health sphere where bone loss is a major concern after total hip arthroplasty.

First, minor expansion of the periosteal surface can be viewed as both an efficient and effective strategy in the adaptation of adult bone to mechanical loads. This suggests that skeletally mature individuals may still have the ability to functionally adapt to changes in mechanical loading through the slow accrual of bone at the periosteal envelope.

Second, bone loss following surgical implantation of a femoral prosthesis has been a long-standing concern with total hip arthroplasty. Implantation can, in the long term, lead to loss of implant support, implant subsidence, and implant or bony fracture. The observed increase in periosteal area and its significance to the maintenance of biomechanical integrity has been neglected in the field of total joint replacement. This is an important point, as the specific details of changes in mass, architecture, and strength should be taken into consideration when interpreting or comparing clinical data after surgery and in the development and surgical implantation of bone prosthetics.

Last, the small sample size renders these results tentative and, therefore, further analysis of larger clinical samples is needed. In particular, the changes observed here require confirmation by more controlled studies documenting changes in physical activity, diet, medical history, and time since implantation.

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