Alterations in femoral strain following hip resurfacing and total hip replacement

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Bone surface strains were measured in cadaver femora during loading prior to and after resurfacing of the hip and total hip replacement using an uncemented, tapered femoral component. In vitro loading simulated the single-leg stance phase during walking. Strains were measured on the medial and the lateral sides of the proximal aspect and the middiaphysis of the femur. Bone surface strains following femoral resurfacing were similar to those in the native femur, except for proximal shear strains, which were significantly less than those in the native femur. Proximomedial strains following total hip replacement were significantly less than those in the native and the resurfaced femur.

These results are consistent with previous clinical evidence of bone loss after total hip replacement, and provide support for claims of bone preservation after resurfacing arthroplasty of the hip.

Metal-on-metal resurfacing of the hip has reemerged in the last decade as a viable alternative to conventional total hip replacement (THR) in certain cases. [1,2] The relatively high success rate of the latest generation of metalon-metal implants is due largely to improve ments in manufacturing techniques and materials, leading to improvement in the wear properties of the implant surfaces.[3] Survival rates for hip resurfacing in the short term are currently slightly less than those for conventional THR using a porous-coated uncemented tapered stem, [4,5] although the long-term success rates for hip resurfacing are unknown. [6-9] The majority of failures following hip resurfacing are due to fractures of the neck, with an approximate incidence of 2%.[10] Suggested causes of fracture include notching of the femoral neck during surgery, varus placement of the femoral component, or poor quality bone stock in the neck, although the exact mechanism remains unknown. 10-12

Hip resurfacing has several theoretical advantages over conventional THR. The limited resection of bone and preservation of bone stock is desirable in younger patients, who may eventually need one or more revision opera tions.[13,14] The femoral resurfacing component is thought to maintain a loading state similar to that of the native femur, although this claim is largely unsubstantiated. Bone mineral density studies have provided evidence for greater preservation of femoral bone with hip resurfacing compared with conventional THR. 15,16 Implantation of the femoral component reduces loading of the proximal aspect of the femur and leads to post-operative resorption of bone, a condition often referred to as stress shielding.[17-19] A reduction in the loss of bone mineral density in the proximal aspect of the femur has been found with the use of uncemented, tapered femoral components compared to uncemented straight femoral components, although clinical evidence indicates that stress shielding and bone loss still occur with tapered components.[20-22]

Currently, we know of no published experimental studies which measure surface strains in the femur prior to and after hip resurfacing and THR. We hypothesised that a hip resurfacing implant would maintain surface strains similar to those of the native femur, whereas a tapered femoral component would create lower proximal strains consistent with stress shielding of the femur. The aim of this study was to measure bone surface strains in cadaver femora prior to and after implantation of a hip resurfacing fem oral component and implantation of an uncemented, tapered femoral component.

Materials and Methods

Bone surface strains were measured at four sites (proximomedial; proximolateral; disto medial; distolateral) on cadaver femora during application of loads representative of the single-leg stance phase of walking for untreated (native), resurfaced and THR configurations. Three pairs of cadaver femora were studied. Radiographs of the specimens were obtained to screen for femoral abnormalities and to ensure that they were compatible with the sizes of the resurfacing component and the tapered stem. All femora were fresh-frozen and stored at -20°C. They were defrosted overnight. The soft tissues were removed and the bones were wrapped in cotton gauze wet ted with saline solution.

One three-element, 45° stacked rosette strain gauge (model 060WR-350, Vishay Micro-Measurements, Raleigh, North Carolina) and three uniaxial strain gauges (model 125UN-350, Vishay Micro-Measurements, Raleigh, North Carolina) were bonded to the surface of the femur (Fig. 1) using a cyanoacrylate adhesive and catalyst following preparation of the bone surface with light sanding, 100% EtOH and M-prep Neutralizer 5A (Vishay Micro-Measurements, Raleigh, North Carolina). The proximomedial rosette gauge was placed approximately 20 mm distal to the centre of the lesser trochanter on the medial surface of the bone, with the posterior edge of the gauge aligned with the posterior surface of the femur in a medial view with the femoral condyles aligned. This is an area where stress shielding has been demonstrated to occur. [19,23] The proximolateral uniaxial gauge was placed on the lateral surface of the bone, centred between the anterior and posterior borders of the femur when viewed later ally, at the same level as the proximomedial gauge. The distomedial and distolateral uniaxial gauges were placed on the medial and lateral surfaces of the femur 250 mm distal to the medial location of the planned THR resection line. The distal gauges were approximately 100 mm distal to the location of tip of the implanted THR component. The ante rior edge of the distomedial gauge was aligned with the medial border of the femur. The posterior edge of the disto lateral gauge was aligned with the centre of the diaphysis when viewed laterally. No gauges were placed in the calcar region because of the possibility of damage during implantation of the prosthesis. All strain gauges were aligned with the longitudinal axis of the femur.

A custom-made fixture designed to reproduce loading conditions during the single-leg stance phase of walking, as described by McLeish and Charnley,[24] was attached to the cross-head of an Instron 1122 Material Testing Machine (Instron Corporation, Norwood, Massachusetts) and used to apply a distributed joint contact force to the femoral head and an abductor muscle force to the greater trochanter (Fig. 2). In order to allow measurement of the joint contact force, a 1000 lb load cell (MLP-1K, Transducer Techniques, Temec ula, California) was placed between the aluminium test fix ture and the acetabular component, which was constructed of a stainless steel housing and a 32 mm diameter UHMWPE shell (Biomet, Warsaw, Indiana). A turnbuckle and 2 mm diameter braided steel cable were connected to a T-condylar one-third tubular plate (Synthes, Paoli, Pennsylvania) attached to the greater trochanter with six 3.5 mm diameter, 20 mm length cancellous bone screws in order to simulate the magnitude and direction of the abductor muscle force. The distal end of the femur was potted in polymethylmethacrylate (PMMA) cement and rigidly attached to a metal wedge at 22° from the horizontal in the frontal plane to reproduce the direction of the joint contact force during slow walking, as described by McLeish and Charnley.[24] The potted femur was positioned so that its frontal plane was parallel with the frontal surface of the proximal test fixture plate. The proximal test fixture was then positioned so that the acetabular component rested on the head of the femur, with the plate of the proximal test fixture in an approximately horizontal position. The metal wedge was then rig idly attached to the material testing load frame.

Prior to testing, the abductor cable was pre-tensioned using the turnbuckle until a second load cell mounted in series with the cross-head of the material testing system measured 20 N in compression, at which point all strain gauge offsets were zeroed. A compressive load up to a max imum of 600 N was then applied through the test fixture with the material testing system under displacement control at a rate of 40 mm/min. A pin passing through the proximal test fixture allowed it to rotate in the frontal plane of the femur. The resulting femoral surface strains and joint con tact force were recorded using an IOtech DBK43A data acquisition system (IOtech, Cleveland, Ohio) at a rate of 10 Hz. Abductor tensions were calculated using the speci men-fixture geometry, measured cross-head load and joint contact force. The load magnitude was limited to 600 N in order to reduce the risk of a femoral fracture and/or abduc tor plate screw pullout. A repeated-measures design exper iment was performed to control the differences between cadaver femora, including variations in femoral bone density amongst the specimens.

Following testing of the native femur, the femoral head was resurfaced using the Articular Surface

Replacement (ASR) implant (DePuy Orthopaedics Inc., Warsaw, Indiana) with a 49 mm diameter femoral component applied to the prepared femoral head (Fig. 3) with a press-fit technique according to the surgical procedure recom mended by the implant manufacturer (DePuy ASR Surgical Technique, Cat. No. 9998-02-280). Care was taken to avoid varus positioning of the component and to avoid notching of the femoral neck. All the procedures were per formed by the same experienced attending joint replace ment surgeon (AJ). Strain gauges remained in the same positions as during testing of the native femur. Testing was then repeated for the resurfaced femur under the same test conditions to which the native femur was subjected.

Next, an osteotomy of the femoral neck was performed approximately 1.5 cm proximal to the lesser trochanter. The femur was then broached to accommodate a size 11 x 142 mm uncemented titanium tapered femoral component (Taperloc, Biomet Orthopedics Inc., Warsaw, Indiana) according to the surgical procedure recommended by the manufacturer (Taperloc Hip System Surgical Technique, Form No. Y-BMT 745/022802/K). The stem was press-fitted into the femur, and a 32 mm cobalt-chrome head (Taperloc) was placed on the stem (Fig. 4). Testing was then repeated for the femur implanted with the tapered component.

All the specimens were tested at room temperature and wetted with saline during testing. Each test for each treat ment was performed three times in order to ensure repeat ability, without any delay between the tests. Data from the third test were used in the analyses. Comparisons of strains were made at a joint contact force of 1100 N for consis tency, including principal strains that were calculated for the rosette gauge. Data from three pairs of femora were used in the analyses, and strain values from the left and right femora of each pair were averaged for an effective sample size of three subjects. A repeated-measures analysis of variance (ANOVA) was performed using Statview Soft ware (SAS Institute, Cary, North Carolina) to determine whether there were significant differences in bone surface strains, joint contact forces and abductor tensions between the native, resurfaced and THR femur configurations ($p < 0.05$, significant). *Post hoc* Fisher's protected least-squares difference tests (Statview software, SAS Institute) were performed to detect pairwise differences in strain between the native, resurfaced and THR configurations.

Results

At a material testing system load of 600 N the mean joint contact force was not different between femur configura tions (1212 N (SD 97)), native; 1276 N (SD 23), resurfaced; 1278 N (SD 62), THR. The maximum abductor tensions were also not statistically different between configurations (615 N (SD 98), native; 679 N (SD 23), resurfaced; 680 N (SD 62), THR.

Bone strains were compared at a standard joint contact force of 1100 N (Fig. 5). For the proximomedial site, max imum principal strains were compressive and approximately aligned with the longitudinal axis of the femur. Differences in mean principal strain angles between the femur configurations were less than 4° and not statistically significant. The mean magnitude of the proximomedial compressive principal strain for the native configuration was 621 (SD 96) microstrain. After resurfacing, strains were not significantly different from those of the native configuration. However, implantation of a tapered femoral component resulted in a 28% reduction in compressive principal strain compared to native ($p \leq 0.007$) and an 18% reduction compared to the resurfaced configuration ($p = 0.037$). Mean tensile principal strains in this location ranged from 140 (SD 13) microstrain after resurfacing to 160 (SD 7) microstrain for native, and were not significantly different between configurations. Mean maximum shear strain at the proximomedial site was 391 (SD 50) microstrain for the native configuration. After resurfacing there was a 12% reduction in maximum shear strain ($p \leq 0.008$), and implantation of a tapered component created a 24% reduction in shear strain from the native ($p \leq 0.001$). The mean maximum shear strain was also different between the resurfaced and THR configurations ($p \leq 0.006$).

For the proximolateral site, maximum axial strains were tensile, with a mean strain magnitude of 637

(SD 59) microstrain for the native configuration (Fig. 5). The 21% reduction in mean strain magnitude for the THR configu ration compared to native was marginally insignificant (p = 0.068). Resurfaced and THR strains were not statisti cally different at this site.

For the distomedial and distolateral sites strains were somewhat variable and less than those recorded in the proximal aspect of the femur (Fig. 5). The mean axial strains at the distomedial site were tensile, with values rang ing from 40 (SD 35) microstrain for the native configuration to 144 (SD 132) microstrain after resurfacing. The SDs were large, particularly for the resurfaced and THR configurations. None of the differences in strains were found to be statistically significant between configurations. The mean axial strains at the distolateral site were found to be compressive, with values ranging from 126 (SD 57) microstrain for the native configuration to 232 (SD 145) microstrain for the resurfaced configuration. The strain differences were not statistically significant between configurations.

Discussion

We hypothesised that a femur implanted with a hip resurfacing component would preserve the physiological strain state found in the native femur compared to a femur implanted with an uncemented, tapered component. Clinically, the con sequences of strain (or stress) shielding are not entirely clear, although stress shielding following THR with uncemented stems is often cited in the literature as a reason for concern. 25 Several long-term clinical studies have demonstrated that there are no known complications directly related to stress shielding following THR .[26-28] However, it is generally agreed that maintenance of bone in the proximal aspect of the femur is desirable, and is of particular importance if the need for revision arises. Younger patients who are candidates for hip resurfacing are more likely to eventually need a revision, and preservation of femoral bone stock to support an implant during revision becomes especially important. [14] Our results in this cadaver model demonstrate that there is a significant reduction in strain from the native state in the proximo-medial aspect of the femur following implantation of a tapered component. However, with the exception of maxi mum shear strains, mean strains measured in the proximal aspect of the femur following implantation of a femoral resurfacing component were not found to be significantly different from strains measured in the native femur.

Reductions in the strain measured in the proximomedial aspect of the femur following implantation of an uncemented, tapered component in the current study are consistent with areas of bone resorption observed in clinical studies of bone mineral density changes following THR using an uncemented, tapered component.[29-32] Many clini cal studies have demonstrated that the largest loss in bone mineral density following implantation of an uncemented, tapered component occurs in the proximomedial aspect of the femur. [29-32] Several of these studies have also provided evidence that a significant amount of bone loss may occur post-operatively in the proximolateral aspect of the femur. [29,30,32] However, Dan et al[31] did not find statistically significant reductions in bone mineral density in the proximolateral aspect of the femur following implantation of an uncemented, tapered component at one year after operation. The high value of SD for strain levels measured in the proximolateral aspect of the femur following implanta tion of the uncemented, tapered component used in this study suggests that stress shielding may be occurring in this region for some individuals, whereas in others the magni tudes of loading approach those of the native femur.

The low values of strain measured in the distal aspect of the femur are consistent with the findings of Oh and Harris [33] and Otani, Whiteside and White [34] that longitudi nal strain decreases from proximal to distal on the surface of the native femur *in vitro*. Additionally, Otani et al[34] found that strains in the distal aspect of the femur became compressive on the lateral surface under axial loading, which is consistent with the findings of our study. This is in disagreement with Kim et al,[18] who found that strains increased from proximal to distal in the intact femur, and Jasty et al,[35] who found that strains were relatively uniform along the diaphysis of the femur. In the studies of Kim et al[18] and Jasty et al[35] the distal end of the femur was con strained using a pinned support, allowing the femur to rotate freely about its anteroposterior axis. Deflections of the mid-diaphysis and distal aspect of the femur under loading would be substantially different from those of a femur which was potted distally and constrained in rota tion, as in our experiment. As a result, the type of fixture used to support the femur distally may have more of an effect on strain magnitudes in the distal aspect of the femur than the type of implant when comparing between studies.

Our study had several strengths, including the use of a cadaver model with simulated abductor muscle loading and attachment of a load cell to the acetabular component to allow precise measurement of forces acting on the head of the femur. The use of paired femora and a repeated-measures design helped to control for differences in femoral bone density among the specimens, which is often a chal lenging problem when testing cadaver specimens. Addition ally, pre-screening of the femora allowed for the use of one size of resurfacing and tapered components, eliminating any effects that could potentially be created by the use of different sizes.

There were also several limitations to our study. The fem oral component was tested in its immediate post-operative condition, in which there is no bone ingrowth into the proximally porous-coated implant. Bone ingrowth would allow for greater load transfer to the cortex of the proximal aspect of the femur and a potential increase in strain values measured in this region. Although the repeated-measures design provided some control over differences in bone min eral density, randomisation of the order of testing implants was not possible. However, load magnitudes were rela tively low compared with *in vivo* levels, [36] and strain values measured during the three consecutive tests of each implant were highly repeatable, indicating that there was little or no degradation of bone properties caused by repeated testing. Despite limited loading magnitudes, some subsidence of the femoral components was observed during testing. Subsid ence may have had an effect on stress shielding levels mea sured in the proximal aspect of the femur for the THR component. Additionally, differences in the head/neck length of the implanted femur from the native femur may have created variations in loading conditions during test ing. However, the mean joint contact forces were not statistically different between the three testing configura tions, indicating that differences in loading conditions were minimal. Another limitation to our study was the lack of a complete set of muscle forces owing to the difficulty of sim ulating muscles around the hip and femur in a laboratory setting. Whereas the hip joint contact and abductor muscle forces have been demonstrated to have the greatest effect on strain patterns in the proximal aspect of the femur, [37] the quadriceps muscles and iliotibial band have been shown to have more of an effect on diaphyseal strains than proximal strains. [38,39] The addition of these muscle groups could pos sibly assist in stabilising the femur, and may help to explain the high SDs in strain measured in the distal aspect of the femur during testing.

To the best of our knowledge there are currently no pub lished experimental studies in which strain patterns were mea sured in cadaver femora in the native state, followed by both hip resurfacing and conventional THR procedures. There are a few published finite-element analysis studies of strain changes in the proximal aspect of the femur following metalon-metal hip resurfacing, but these were limited to examina tion of the femoral head and neck .[40,41] Kishida et al[15] performed a comparative clinical study demonstrating sig nificantly greater loss of bone mineral density in the proxi mal femur for a cementless stem compared to a metal-on metal hip resurfacing prosthesis using dual-energy X-ray absorptiometry. Harty et al[16] similarly found that the bone mineral density was maintained in the proximal aspect of the femur for a metal-on-metal resurfacing implant. Our results complement the findings of these studies and dem onstrate that strain patterns in the femur following hip resurfacing are similar to native patterns, although implan tation of a resurfacing component may result in reductions in shear strains in the proximomedial aspect of the femur. Implantation of an uncemented, tapered, femoral compo nent results in significantly larger reductions in strains from the native state in the proximomedial aspect of the femur compared to resurfacing, leading to a greater potential for bone loss in this region.

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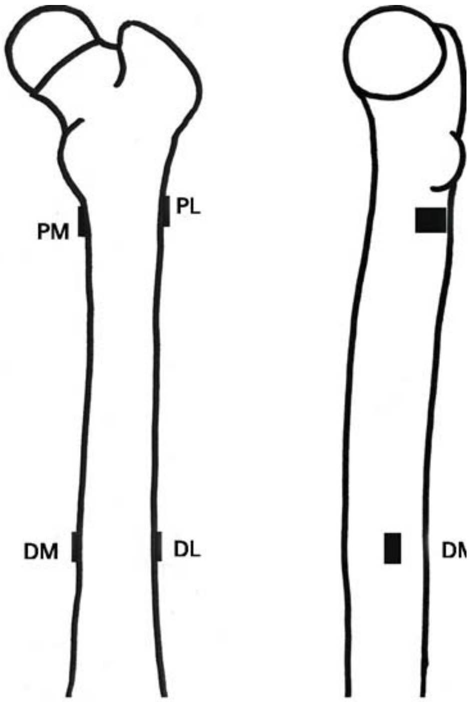

Fig. 1

Drawing showing posterior (left) and medial (right) views of a resurfaced right femur showing strain gauge locations on the proximomedial (PM), proximolateral (PL), distomedial (DM), and distolateral (DL) surfaces.

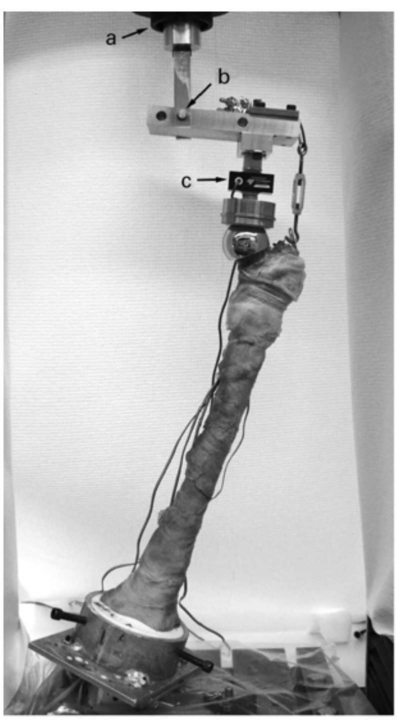

Fig. 2

Photograph showing the femoral test fixture used to simulate loading conditions during the single-leg stance phase of walking as described by McLeish and Charnley.21 A 600 N compressive load was applied by the cross-head through a pin (a) and to the proximal test fixture (b). The joint contact force was measured using a load cell (c) mounted between the acetabular component and the test fixture.

Fig. 4

Photograph showing a left femur implanted with an 11 mm taperloc stem and a 32 mm head.

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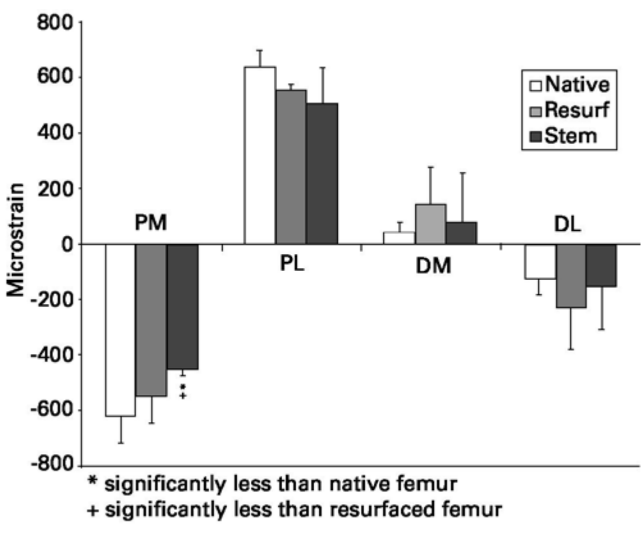

Fig. 5

Graph showing femoral strains at a hip joint force of 1100 N at four locations for the native, resurfaced (resurf), and tapered stem (stem) configurations. Strains in the proximomedial aspect of the femur implanted with a tapered stem were significantly less than in the native femur and the resurfaced femur (PM, proximomedial; PL, proximolateral; DM, distomedial; DL distolateral).