RESEARCH

Increased strain in the femoral neck following insertion of a resurfacing femoral prosthesis

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The cortical strains on the femoral neck and proximal femur were measured before and after implantation of a resurfacing femoral component in 13 femurs from human cadavers. These were loaded into a hip simulator for single-leg stance and stair-climbing. After resurfacing, the mean tensile strain increased by 15% (95% confidence interval (CI) 6 to 24, \( p = 0.003 \)) on the lateral femoral neck and the mean compressive strain increased by 11% (95% CI 5 to 17, \( p = 0.002 \)) on the medial femoral neck during stimulation of single-leg stance. On the proximal femur the deformation pattern remained similar to that of the unoperated femurs.

The small increase of strains in the neck area alone would probably not be sufficient to cause fracture of the neck. However, with patient-related and surgical factors these strain changes may contribute to the risk of early periprosthetic fracture.

Resurfacing arthroplasty of the hip has re-emerged as an option in total hip replacement (THR). In 2008 it constituted 7.8% of all primary hip replacements in Australia.\(^1\) In Scandinavia, however, the use of resurfacing prostheses is substantially less, recorded as between 0.6% and 2.8% in the national arthroplasty registries.\(^2,4\) The medium-term clinical follow-up shows promising results for the recent generations of resurfacing implants.\(^5,11\)

Their obvious advantage is the preservation of proximal bone stock, which simplifies revision procedures. These prostheses retain the anatomical joint and a physiological transfer of load is expected in the proximal femur. Periprosthetic fractures of the neck, with an incidence reported from 0.8% to 2.5%,\(^5,9,12-14\) is the most frequent complication, followed by aseptic loosening.\(^11,15\) Periprosthetic fractures after resurfacing arthroplasty are associated with patient characteristics such as female gender,\(^12,13,16\) age,\(^13,16\) obesity\(^12-14\) and femoral abnormalities.\(^8,14\) Careful selection of the patients can reduce the risk of fracture.\(^8\) Surgical factors such as varus positioning of the implants,\(^13,17\) notching\(^13,18\) and lack of seating\(^18\) are also associated with an increased risk of fracture.

Periprosthetic fracture of the neck occurs early after resurfacing arthroplasty, typically around two months after operation.\(^15\) Within a few months after the resurfacing arthroplasty there appears to be apposition of bone in the femoral neck,\(^19-22\) which may reduce the risk of fractures in the long term.

Finite element analyses (FEA) and experimental studies have demonstrated increased strain in the femoral neck after implantation of a resurfacing prosthesis.\(^20,23-27\) This has been related to the risk of fracture, but the evidence is unclear. A recent experimental study found both stress concentration and stress shielding in the femoral neck following implantation of a resurfacing component, depending on the positioning of gauges.\(^28\) Vail et al.\(^29\) measured the pattern of strain using a photoelastic method on cadaver femurs and showed reduced strain in the neck when the implant was in a neutral position, in contrast to most other studies. The findings in numerical and experimental studies on the load pattern after resurfacing arthroplasty are thus confusing.

The aim of this study was to measure the strains on the femoral neck and proximal femur in human femurs from cadavers before and after implanting a femoral resurfacing component.

Materials and Methods

There were 13 human femurs, collected from five males and eight females whose mean age was 62.1 years (44 to 75) (Table I). Radiographs were taken of each femur to exclude localised skeletal abnormalities. The bone mineral density (BMD) in the neck and trochanteric areas was obtained by dual-energy X-ray absorptiometry (Hologic Discovery A, Bedford, Massachusetts, Table I). Femurs with T-scores below -2.5 for the total
proximal femur were considered osteoporotic and excluded from the study which was approved by the regional medical research ethics committee.

The femurs were prepared and tested according to a method previously described.\textsuperscript{30} They were wrapped in wet towels in plastic bags and stored at -20°C. Before testing, they were thawed at room temperature and any remaining soft tissues removed. Before they were resected, the femoral condyles were used as a reference to establish frontal and sagittal planes. The femurs were mounted in a metal cylinder and fixed with bone cement (Meliodent, Heraeus, Hanau, Germany). The proximal 25 cm, measured from the tip of the greater trochanter, were kept above the cylinder. A nylon strap was attached to the greater trochanter with five screws to simulate the abductor muscles (Fig. 1).

A total of ten pre-wired strain gauge rosettes (Tokyo Sokki Kenkyujo Co. Ltd, Tokyo, Japan) were glued (X60, HBM, Darmstadt, Germany) to the femur in predetermined positions (Fig. 2). Each rosette comprised three strain gauges at an angle of 45° apart. The rosettes were orientated parallel to the longitudinal axis of the femur. Seven were distributed at three levels on the anterior, posterior and medial aspects of the proximal part of the shaft. In the neck region three were attached at the lateral, medial and anterior aspects, respectively. They were positioned over the centre axis of the neck in the frontal and lateral planes. The gauge outputs were recorded by a measurement amplifier (UPM 100, HBM).

The resurfacing prostheses (DePuy ASR, DePuy International Ltd, Leeds, United Kingdom) were implanted by the senior author (AA) according to the manufacturer’s recommended technique (Fig. 3). They were aligned between neutral and 10° valgus relative to the centre axis of the neck and fixed with a thin layer of cement (SmartSet HV Cement, DePuy, International Ltd), taking care to avoid leaving cement around the central pin.

Under identical set-up and loading conditions, the cortical strains were first measured on the intact and then on the operated femur. The femur was mounted in a custom-made hip jig, then placed in a material testing machine (MTS 858

\begin{table}
\centering
\caption{Specimen characteristics. Identification (ID) number given in the laboratory. L and R indicate left and right, respectively. The prosthesis size is based on the bearing diameter of the femoral component.}
\begin{tabular}{llllll}
\hline
ID number & Gender & Age (yrs) & Total bone mineral density (g/cm$^2$) & T-score & Prosthesis size (mm) \\
\hline
1R & M & 59 & 0.89 & -0.9 & 53 \\
2L & F & 74 & 0.75 & -1.5 & 47 \\
3L & F & 61 & 0.83 & -0.9 & 46 \\
6L & F & 58 & 0.65 & -2.4 & 49 \\
8L & M & 49 & 0.87 & -1.1 & 53 \\
10R & F & 70 & 0.71 & -1.9 & 49 \\
11R & F & 69 & 0.81 & -1.1 & 45 \\
12R & F & 62 & 0.98 & 0.3 & 46 \\
13L & M & 74 & 1.04 & 0.0 & 53 \\
15R & M & 63 & 0.83 & -1.5 & 49 \\
16L & F & 64 & 1.04 & 0.8 & 46 \\
26L & M & 60 & 1.08 & 0.3 & 53 \\
29R & F & 44 & 0.80 & -1.2 & 49 \\
\hline
\end{tabular}
\end{table}
MiniBionix II, MTS Systems Corporation, Eden Prairie, Minnesota), aligned in 12° of valgus, corresponding to the physiological inclination of the leg in the stance phase.\textsuperscript{31} The centre of the acetabular component was positioned 11 cm lateral to the load axis, in accordance with the geometry of an average pelvis. Load was applied to the femoral head by a lever arm connected to the piston of the testing machine. In order to avoid unphysiological bending, the femur was free to rotate around its longitudinal axis and to tilt in the medio-lateral plane. Stair-climbing was simulated by applying torsional load to the distal femur by a transverse crossbar connected to the metal cylinder holding the specimen. The trochanteric strap simulating the abductor muscles was held at an angle of 15° to the longitudinal axis of the femur. In order to simulate the force of the iliotibial track, a wire extension from the trochanteric strap ran over two pulleys mounted on the outer end of the lever arm (Fig. 1). The pulleys were assumed to be friction free and the force on the femur was distributed equally in the trochanteric strap and iliotibial band. The forces in the iliotibial band were monitored by a 5 kN load cell (U9B, HBM). Another 2 kN load cell (U9B, HBM) recorded the anteroposterior force on the femoral head. The median resultant force on the hip joint for each configuration was calculated based on the vertical force and the forces recorded by the load cells.

On the pelvic side of the jig, the intact femoral head articulated with an appropriately sized acetabular component (DePuy ASR) aligned at 45°. If necessary, this was changed after surgery to the size corresponding to the femoral component.

All the femurs were subjected to four loading configurations. We applied two vertical forces, 600 N and 900 N, to simulate single-leg stance, and to simulate stair-climbing we added 10 Nm and 15 Nm torsional forces, respectively. In each case the measurements were repeated three times and the mean values used for comparison of the principal strains.\textsuperscript{32} Because of the physiological bending moment, we chose to investigate the principal compressive strains on the medial surface of the femur and principal tensile strains on the lateral and anterior aspects.

Statistical analysis. Statistical analysis was performed using the software package SPSS 14.1 (SPSS Inc., Chicago, Illinois). The mean change in strain between the intact and the operated femurs was calculated as the percentage of intact strain values, reducing the impact of variation between the different femurs. The mean change in strain was assessed at the position of each strain gauge by comparing the percentage of the intact value with 100%, analysed by one-sample \textit{t}-test. A significance level of \( p < 0.01 \) was chosen to avoid errors caused by multiple comparisons, as there was a total of ten gauge positions. The percentage changes of strain were normally distributed.

A power analysis of the primary research question of mean change in strain analysed by one-sample \textit{t}-test was performed with a significance level of 0.05 and power of 80%. A total of 13 specimens were sufficient to detect differences of at least a 25\% change in strain on the lateral side, SD 45\%, and of at least 15\% on the medial side, SD 46\%.

We also investigated by multiple linear regression the influence of the co-factors of age and BMD on the change in strains in the neck region expressed as the percentage of intact strain. A significance level of \( p < 0.05 \) was chosen for this analysis. The assumptions underlying the regression analysis were checked by a study of the residuals of...
change in strain expressed as the percentage of intact strain. The residuals were normally distributed with a constant variance.

The differences in the calculated resultant force on the hip joint had a skewed distribution, and median and interquartile ranges were used to present these results. Comparison of forces before and after surgery were analysed by Wilcoxon’s signed ranks test with a significance level of \( p < 0.05 \).

**Results**

After resurfacing, the main result of this study was a 14\% (95\% confidence interval (CI) 6 to 23, \( p = 0.004 \)) and 15\% (95\% CI 6 to 24, \( p = 0.003 \)) increase in the mean principal tensile strain on the lateral aspect of the femoral neck following 600 N and 900 N single-leg stance simulation, respectively. Correspondingly, a 10\% (95\% CI 4 to 16, \( p = 0.003 \)) and 11\% (95\% CI 5 to 17, \( p = 0.002 \)) increase in the mean principal compressive strain was measured on the medial aspect (Fig. 4a).

For stair-climbing, the increase in the mean strain was clearly less than for single-leg stance on the medial and lateral aspects of the neck (Fig. 4b). The greatest increase in strain was on the anterior aspect and for a 600 N vertical force the mean increase was 16\% (95\% CI 2 to 30, \( p = 0.03 \)), which was not statistically significant.

The largest principal strain values were observed with the 900 N vertical force with a mean compressive strain of 2089 microstrains (\( \mu \varepsilon \)) (95\% CI 1663 to 2515 \( \mu \varepsilon \)) as the largest value seen on the neck.

In the proximal femur the strain measurement revealed a deformation pattern on the operative femurs similar to the intact femurs for both 600 N and 900 N loading levels (Fig. 4).

The results from multiple linear regression of co-factors suggest that increasing age by ten years would imply a 7\% to 11\% increase in strain medially and laterally in the femoral neck, after adjusting for the effect of BMD. Furthermore, an increase of BMD by 0.1 g/cm² would give an average 7\% reduction in strain laterally during stair-climbing, after adjusting for the effect of age.

For single-leg stance there was a statistically significant reduction in the median resultant hip joint force of 37 N for 600 N vertical force (\( p = 0.01 \)). For stair-climbing there was a statistically significant reduction of 19 N for 600 N vertical force (\( p = 0.04 \)).

**Discussion**

In this study we have demonstrated that implantation of a resurfacing femoral component increases strain in the area of the femoral neck. During single-leg stance, the mean compressive strains on the medial side and the mean tensile strains on the lateral side were significantly increased compared to the intact femur. During stair-climbing the mean strains increased on the anterior side, but this was not statistically significant.

An increase in cortical strain on the femoral neck following hip resurfacing has been shown in an experimental study on composite femurs, and predicted by finite element analysis studies. However, a recent investigation found generally small alterations in strains after implantation of a resurfacing femoral component, showing some localised regions of peak stress concentration and other regions of shielding. In contrast in earlier studies, using photoelastic strain measurement in cadavers, Vail et al. found reduced strains in the neck after resurfacing. As measurement by strain gauge is considered a more accurate method, and the present study included a relatively large sample size, we believe that we can confirm that strains increase in the neck after implantation of a resurfacing femoral component. Based on these findings, stress shielding would not be expected in the femoral neck area after resurfacing arthroplasty. Rather, increased stress would lead to bone formation within a few months as shown by FEA and DEXA, and some investigators suggest that this would reduce the risk of fracture in the long term. The contribution of increased strain to the risk of early fracture of the neck is uncertain. Some authors have found the increase in stress to be relevant to the initial risk of fracture, but others have described the absolute strain levels as small compared with yield strain, and therefore unlikely to cause fractures. Cristofolini et al. found localised peak stress concentrations which they related to early fractures of the neck. Their findings were, however, based on a small sample, and the relevant results were not significantly different from intact strain values. Our experiment gave a small, but significant increase in strain on the implanted femur, with an increase of up to 15\% on the femoral neck in single-leg stance. The thresholds of strain for critical damage are suggested to be 2500 \( \mu \varepsilon \) and 4000 \( \mu \varepsilon \) for tension and compression, respectively. The mean levels at all our strain gauge positions were well below these limits. The increase in strain alone after resurfacing arthroplasty would therefore probably not be sufficient to cause periprosthetic fracture of the neck.

As expected, we found that the strain pattern on the proximal femur after hip resurfacing resembled the physiological pattern of the intact femur. This was supported by a recent experimental study on cadaver femurs. Thus, stress shielding is not to be expected either in the femoral neck or proximal femur after resurfacing arthroplasty. Also, Deuel et al. showed that an uncemented tapered femoral stem significantly reduces strain in the proximal medial region of the femur compared to both the intact and the resurfaced femur. Strain measurements in our study showed a substantial improvement in the pattern in the proximal femur for the resurfacing prosthesis compared to experiments on standard THR performed earlier in the same laboratory with the identical apparatus. Our findings are in agreement with those of Deuel’s study.

Factors specific to patients influence the result of resurfacing arthroplasty. Several studies emphasise the impor-
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We found that an increase in BMD significantly reduced the mean strain only at the lateral aspect of the neck for stair-climbing, after adjusting for the effect of age. A relationship between a reduction in BMD and increased strain is in agreement with a mechanical study which proved a strong correlation between decreased BMD and reduced fracture strength.37 Furthermore, we found that for single-leg stance, increased age significantly increased the mean strain, both at the medial and lateral aspects of the neck and medially after adjusting for the effect of BMD. The revision rate after resurfacing is reported to be substantially higher for men older than 65 years and women above 55 years,1 with one of the main selection criteria being younger age.13,38 Our findings also support age as a selection criterion. The number of subjects in this study may have been insufficient to detect all relevant influences of the covariates,
as the sample size was adjusted for analysis of change in strain and the regression analysis was supplementary.

Varus alignment of the resurfacing prosthesis has been associated with an increased risk of fracture.\textsuperscript{13,17} In this study, implant position was surgeon-based, aiming for neutral to 10° valgus relative to the centre axis of the neck. The effect of varying the position of the resurfacing component on the load pattern was therefore not covered, but has been considered to some extent in other studies. An increase in strain in the neck area has been demonstrated when orienting the prosthesis in varus,\textsuperscript{29} with a corresponding decrease in strain when it is in valgus.\textsuperscript{39} Cadaver models have shown that the ultimate fracture load increases when the femoral component is in valgus.\textsuperscript{37,40,41} Indeed, some authors advocate maximal valgus of the prosthesis,\textsuperscript{40,41} even though this may have an adverse effect on the fracture load.\textsuperscript{37,41} It is clear that the consequences of extensive valgus orientation are still not fully explored, and further research is warranted.

A limitation of this study was that measurement with the strain gauge reflects deformation of the cortical surface only at the exact position of the strain gauge rosette. The positioning of the gauges is therefore highly relevant. One would expect maximum compression and tension medially and laterally in the neck in single-leg stance. During simulated stair-climbing we observed that the greatest compressive strain on the medial side and tensile strain on the lateral side was reduced compared to single-leg stance, both for intact and operated femurs. This may be explained by the change in direction of the resultant hip force between the two loading circumstances. The position of the strain gauges on the neck may not have been optimal in detecting principal strain values for simulated stair-climbing.

Experiments in vitro are a simplification of the situation in vivo. The set-up for load in a hip simulator should be as simple as possible because the magnitude of the muscle forces is difficult to estimate.\textsuperscript{42} However, the abductor muscles account for at least 50% of the strain changes in the proximal femur and should therefore be included.\textsuperscript{42} It has also been demonstrated experimentally that abandoning the abductor force increases strain values in the neck area.\textsuperscript{25} In this study the iliobial band was also included, as finite element analysis has shown that the difference in principal strain was less than 3% when comparing abductor and iliobial forces combined with a model including all muscles.\textsuperscript{43} Therefore, the simulation of the hip force in this study is considered realistic. Both the measured forces and calculated resultant forces at the hip joint were similar overall both before and after surgery, even though we found statistically significant differences for median resultant hip joint forces in single-leg stance. The largest difference was 37 N with a vertical forces of 600 N, which is approximately 2.5% of the total resultant force on the hip joint. We consider this reduction in the median resultant hip joint force to be small and not a plausible cause for the changes in strain observed in this study. The magnitudes of the resultant hip joint forces are comparable with those of other experimental studies.\textsuperscript{29,35}

and within the range of such forces presented by telemetric studies.\textsuperscript{42,44} Also, the strains in the level of the most distal gauge after resurfacing were within 8% of the strains before implantation for all configurations. According to Cristofolini\textsuperscript{42} and Cristofolini and Viceconti,\textsuperscript{45} such a small variation confirms that consistent loading conditions were applied to the intact and the implanted femur.

We conclude that strain increases in the medial and lateral aspects of the femoral neck after hip resurfacing arthroplasty, whereas a physiological strain pattern is preserved in the proximal femur. The highest observed mean increase in strain in the neck is relatively small, and alone would probably not be sufficient to cause a fracture. However, acting with patient-specific and surgical factors it may contribute to the risk of early periprosthetic fracture.

**Supplementary material**

Tables showing the change of strain from unoperated to operated femur, descriptive results of principal strains for each of the strain gauge rosette positions, results from multiple linear regression analysis of the dependent variable change in strain and median forces recorded in the experimental set up during testing are available with the electronic version of this article on our website at www.jbjs.org.uk

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