*The Open Orthopaedics Journal*, **2011**, *5*, 379-384


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# Differences in External and Internal Cortical Strain with Prosthesis in the Femur

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**Abstract:** The contact between a femoral stem prosthesis and the internal surface of the cortical bone with the stress in the interface is of crucial importance with respect to loosening. However, there are no reports of strain patterns at this site, and the main aim of the current study was to investigate differences of internal and external cortical strain in the proximal femur after insertion of a stem prosthesis. The external cortical strain of a human cadaveric femur was measured with strain gauges before and after implantation of a stem prosthesis. By use of optical fibres embedded longitudinally in the endosteal cortex, deformations at the implant–internal cortex interface could also be measured. The main external deformation during loading of the intact femur occurred as compression of the medial cortex; both at the proximal and distal levels. The direction of the principal strain on the medial and lateral aspects was close to the longitudinal axis of the bone. After resection of the femoral neck and insertion of a stem prosthesis, the changes in external strain values were greatest medially at the proximal level, where the magnitude of deformation in compression was reduced to about half the values measured on the intact specimen. Otherwise, there were rather small changes in external principal strain. However, by comparing vertical strain in the external and internal cortex of the proximal femur, there were great differences in values and patterns at all positions. The transcortical differences in strain varied from compression on one side to distraction on the other and vice versa in some of the positions with a correlation coefficient of 0.07. Our results show that differences exist between the external and internal cortical strain when loading a stem prosthesis. Hence, strain at the internal cortex does not correspond and can not be deducted from measured strain at the external cortex.

**Keywords**: Deformation, femur, hip, optical fibre, prosthesis, strain.

## INTRODUCTION

According to traditional hip biomechanics theory, based on mathematical models and finite element analysis, a bending moment acts on the proximal part of the femur [1-3]. This theory has been validated by *in vitro* [4] and *in vivo* strain gauge measurements [5]. Insertion of a prosthesis has been shown to alter the strain patterns of the proximal femur which may have adverse effects on the dynamic bone remodeling and the ultimate fate of the prosthesis [6, 7]. Hence, the prevailing principles for optimal design of prosthesis are to aim for an implant-bone fit that is close to the physiological strain patterns, especially at the proximal aspects of the femur. Strain patterns have regularly been evaluated at the external cortex, both of the intact femur and after insertion of prosthesis. However, the contact zone between the prosthesis and the femoral bone is on the internal surface of the cortex, and the stress in this interface is of crucial importance with respect to the event of loosening. To our knowledge, there are no reports of the strain patterns at this site. Strain gauges are not appropriate for application at the implant-internal cortex interface, but Optical Bragg grating fibres (OBGF) are widely used in the industry for measurements of strain. Fibre Bragg gratings are diffracting elements printed in the core of an optical fibre. They behave like selective filters which in the fibre core reflect the spectral components of a propagating packet according to the Bragg relation $\lambda = 2n\Lambda$; where $\lambda$ is the wavelength, $n$ being the core mean reflective index and $\Lambda$ the spatial period of the refractive index modulation. If the Fibre Bragg Grating is strained along the fibre axis, $\Lambda$ is changed. As a result, the Bragg wavelength is shifted, which is a measure of strain. Previously, we have shown that Optical Bragg grating fibers are well suited for dynamic measurements of bone strain [8]. On this basis, the present study was undertaken to evaluate the strain patterns at the internal and external cortex of the proximal femur before and after insertion of stem prosthesis.

## MATERIALS AND METHODS

One cadaveric femur was removed within 24 hours from a 64 year old male patient who died from a heart attack, packed in saline-soaked towels and stored for –80 degrees. Before testing, the specimen was thawed at room temperature, and the remaining soft tissues were removed. To fit into our testing system the condyles were removed, and the distal part of the diaphysis was cemented into a steel cylinder. The distance from the upper end of the cylinder to the tip of the greater trochanter was 180 mm.

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For strain gauge measurements 8 rosettes (1-RY91-6, Hottinger Baldwin Messtechnik, Darmstadt, Germany) were externally bonded to the cortex of the femur, 4 proximally at the minor trochanter area and 4 at a level corresponding to the tip of the prosthesis. At each level one rosette was bonded medially, anteriorly, laterally and posteriorly. Before bonding the bone surface was smoothed with 400 grit silicone carbide papers and degreased with acetone. Each rosette was cemented to the bone using a cyanoacrylic adhesive (X 60, Hottinger Baldwin Messtechnik, Darmstadt, Germany). The wires from the rosettes were coated with a protective chemical (SG 250, Hottinger Baldwin Messtechnik, Darmstadt, Germany) and connected to amplifiers (Spider 8, Hottinger Baldwin Messtechnik, Darmstadt, Germany). The miniature rosettes have three strain gauges mounted horizontally and vertically and at 45 degrees angles in between on a polyamide carrier. This design allows calculation of direction and magnitude of surface principal strain. The directions are presented as the counter clockwise angle from the horizontal rosette gauge, ranging from 0 to 180 degrees. After bone preparation and application of the strain gauges, the femur was placed on a jig mounted on a material testing system (2,5 kN single column table-top zicki-Line, Zwick Roell®, Ulm, Germany). The specimen was tilted in a way that corresponded to 12 degrees of valgus in relation to the direct vertical load on the femoral head, which corresponds to the physiological inclination during leg stance (ISO-Norm 7206-4) (Fig. **1**).

**Fig. (1).** Experimental setup with strain gauges during mechanical testing. Uncemented femoral stem prosthesis in the medullary canal. Loading of the specimen was performed with the femoral head component applied on the neck of the prosthesis.

To avoid zero shift (hysteresis) and secure stable strain initial gauge readings, the gauges were subjected to a preconditioning process before data acquisition by first preloading the femur with 1000 N axial force, which corresponds approximately to the gravitational load force from the body weight of an adult male, then unloading and taring the strain gauges to zero. Thereafter 1000 N axial load was applied and strain gauge measurements on the intact femur were recorded with a sampling rate of 50 Hz during the middle 3 seconds of the loading cycle while the load was maintained at 1000 N. The minimum and maximum principal surface strain and corresponding principal direction were calculated through strain transformation [5]. Determination of principal direction in degrees from the strain gauges was done by defining the horizontal gauge of each rosette as the zero degree axis. Next, an osteotomy of the neck was performed, and the femoral canal was reamed as recommended by the supplier of the prosthesis. Reaming was carried out for insertion of an uncemented straight stem titanium ($TiAl_6V_4$) prosthesis #9 (Landos Corail, Landanger, Chaumont, France). The outer surface of the prosthesis was sandblasted. We used a 28 mm head with standard neck and offset (Fig. **1**). The prosthesis was preloaded with 1000 N axial load and then unloaded before a new load of 1000 N was applied for strain gauge measurements. It was then removed, and the femoral canal was reamed to fit an uncemented prosthesis #10. Thereafter the same loading and measurements procedures were repeated.

The stem prosthesis was removed and the femoral specimen was prepared with 4 shallow vertical ditches made at the proximal level in the medullary canal at the endosteal cortex corresponding to the position and longitudinal direction of the rosette vertical strain gauges on the outside periosteal cortex. In the small ditches fibre-optic sensors with a diameter of 125 µm and 1550 nm wavelength (Optolink Corporation Ltd., Washington, Tyne and Wear, England) were sealed with the cyanoacrylic adhesive (HBM X 60, Darmstadt, Germany) in a way that they not were engaged by the stem of the prosthesis (Fig. **2**). A fiber Bragg grating analyzer (LightStructures, Oslo, Norway) with a sensitivity of 1.2 pm/microstrain was used. The fiber signals were sampled synchronously with those from the strain gauges and the sampling rate was 60 Hz.

The femoral specimen was then remounted in the jig, and the prosthesis #10 was inserted in an uncemented setting. It was preloaded with 1000 N and then unloaded before new loading and strain measurements were performed as previously described including endosteal fiberoptic signals at the proximal level. The prosthesis was then removed, bone cement was inserted into the medullary canal, and the #9 prosthesis was cemented for new measurement procedures. The difference between #9 and #10 allowed 1 mm space for the cement around the #9 prosthesis (Fig. **3**).

The change in strain recorded by the 4 longitudinally placed optical fibres at the internal cortex was compared to the simultaneous change in strain recorded by the gauges on the external cortex, and the association was tested by use of the Pearson product moment correlation coefficient. Each series was repeated twice. The standard deviation of the 3 measurements for both types of sensors varied from 13 to 49 µm/m; i.e. percent deviation from 1.0 to 5.2%.

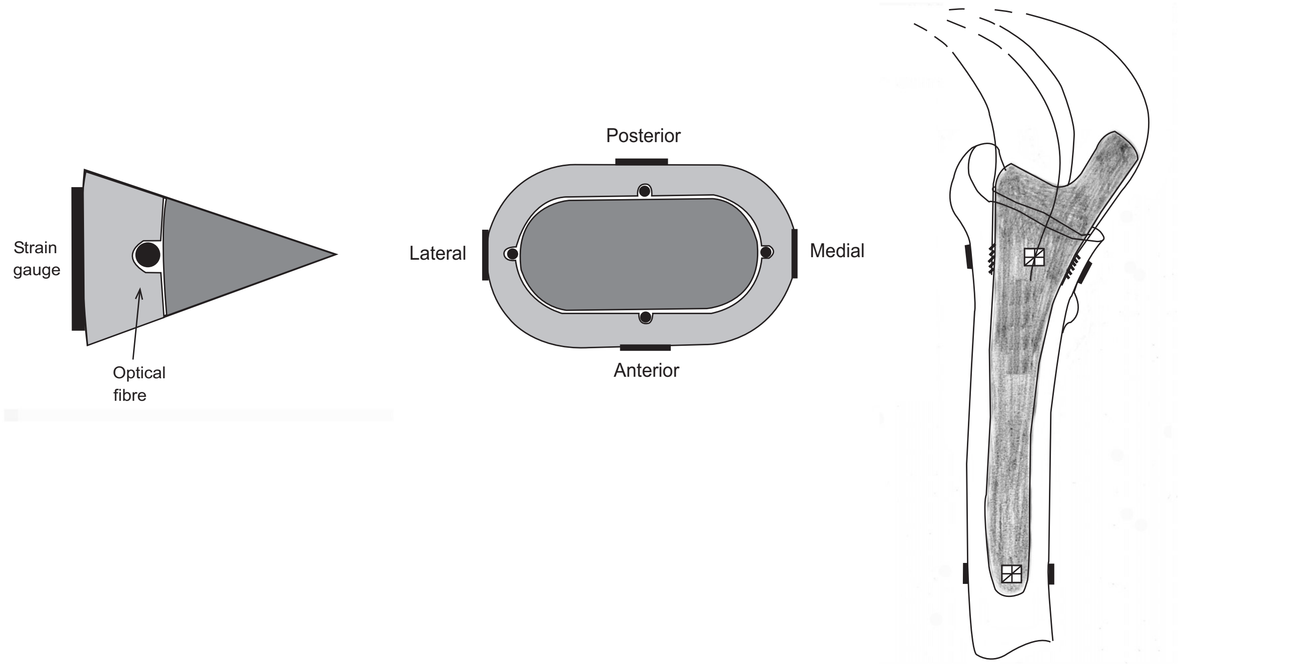


**Fig. (2).** Schematic drawings of the experimental set up with 4 strain gauges at the external femoral cortex both proximally and distally and 4 optical fibers at the internal femoral cortex proximally.


**Fig. (3).** Experimental setup with strain gauges and optical fibers (indicated by arrows). Femoral stem prosthesis cemented in the medullary canal. Loading of the specimen was performed with the femoral head component applied on the neck of the prosthesis.

## RESULTS

Table **1** gives the principal external cortical strain (mean) at the proximal and distal levels of the intact femoral specimen and after neck resection and implantation of thinner and thicker titanium uncemented stem prosthesis. The main deformation during loading of the intact femur was compression of the medial cortex both at the proximal and distal level. The direction of the principal strain on the medial and lateral cortex was close to the longitudinal axis of the bone. After resection of the femoral neck and insertion of a thin uncemented prosthesis, the changes in strain values were greatest medially at the proximal level, where the deformation in compression was reduced to the half. Otherwise, there were rather small changes (less than 10% of the medial change) in principal strain. Change from the thin #9 uncemented prosthesis to a thicker #10 resulted in a 26% reduction of medial compressive strain, else the differences were less significant. Introduction of cement in the medullary canal in combination with a thin #9 prosthetic stem versus the previous uncemented #10 stem situation demonstrated a relative transfer of the compressive strain on the proximal external cortex from medial to anterior.

The external cortex (gauges) and internal cortex (fiber optical wires) vertical strain values and patterns at the proximal level with an implanted uncemented #10 and cemented #9 Titanium stem prosthesis are given in Table **2**. It can be seen that also for the internal cortex optical fiber measurements, as well as for the external cortex, the strain patterns change from the uncemented to the cemented setting of the prosthesis. However, in both settings, there are great differences in strains in all positions from the external to the internal cortex of the proximal femur. The differences in strain varied from compression on one side of the bone to distraction on the other and vice versa in some of the

**Table 1. Principal Strains (Microstrain), Measured by Rosette Strain Gauges at the External Cortex, at the Proximal and Distal Levels of the Femoral Specimen. Orientation Angle of the Principal Strain is Positive Counterclockwise and Relative to the Horizontal Triplet of the Rosette Gauge**

| | | Intact Bone | Uncemented #9 Prosthesis | Uncemented #10 Prosthesis | Cemented #9 Prosthesis |
|---|---|---|---|---|---|
| **Proximal** | Strain (orientation) | | | | |
| Medial | Maximum | 496 (28°) | -82 (179°) | -45 (172°) | -32 (175°) |
| | Minimum | -1096 (118°) | -428 (89°) | -317 (82°) | -384 (85°) |
| | | | | | |
| Anterior | Max | -13 (64°) | 208 (29°) | 124 (28°) | 73 (10°) |
| | Min | -147 (154°)° | -196 (119°) | -161 (118°) | -269 (100°) |
| | | | | | |
| Lateral | Max | 54 (102°) | 49 (103°) | 59 (106°) | 59 (102°) |
| | Min | -32 (12°) | -31 (13°) | -41 (16°) | -38 (12°) |
| | | | | | |
| Posterior | Max | 196 (126°) | 190 (126°) | 153 (124°) | 190 (129°) |
| | Min | -151 (36°)° | -141 (36°) | -123 (34°) | -195 (39°)° |
| **Distal** | | | | | |
| Medial | Max | 115 (178°) | 130 (177°) | 132 (176°) | 126 (177°) |
| | Min | -491 (88°) | -501 (87°) | -477 (86°) | -466 (87°) |
| | | | | | |
| Anterior | Max | 83 (163°) | 78 (164°) | 71 (166°) | 103 (164°) |
| | Min | -183 (73°) | -230 (74°) | -196 (76°) | -230 (74°) |
| | | | | | |
| Lateral | Max | 201 (94°) | 176 (95°) | 214 (97°) | 172 (96°) |
| | Min | -76 (4°) | -29 (5°) | -34 (7°) | -64 (6°) |
| | | | | | |
| Posterior | Max | 62 (44°) | 50 (75°) | 20 (89°) | 89 (61°) |
| | Min | -79 (134°) | -32 (165°) | -36 (179°) | -55 (151°) |

positions with a correlation coefficient of 0.07 for measurements.

## DISCUSSION

The gold standard of measuring bone deformation under load is by application of the strain gauge, a device whose electrical resistance varies in proportion to the amount of strain. However, the use of strain gauges in humans are limited, mostly because they are difficult to adhere to bone, and the strain gauge itself and the measurement wires represent a substantial contaminant. In addition to strain gauges on the external cortex we used in the present study OBGF to measure internal cortical strain in the proximal femur after insertion of a stem prosthesis. A functional optical sensor system is a combination of a light source, sensors and an analyzer that receives the optical signals from the sensors and converts them to a format suited for digital signal processing. We found that external cortical deformation when loading is influenced by the size of the stem whether it is cemented or not, and that external cortical strain is different from internal.

A limitation of our experiment is that we have used only one human specimen and there is great variability of mechanical properties of the proximal femoral bone. However, it was not our aim to investigate such variability, but merely to investigate principal differences between strain measure on the outer cortical region as compared to the inner cortical region of the proximal femoral bone, without and with a stem prosthesis. Principally, this can be verified by one femoral specimen.

A structure monitoring system using fiber-optic sensors is based on OBGFs. The gratings are short sections of optical fiber that have been sensitized to strain and temperature through a periodic modulation of the refractive index of the fiber core. A grating reflects a narrow range of wavelengths (colours) determined by the period of the modulation. The most strongly reflected wavelength is called the Bragg wavelength and is a characteristic of the grating. The period of the grating is commonly chosen such that the reflected wavelength falls in the parts of the infrared spectrum used by the telecommunications industry, i.e. around 1300 or 1550 nm. Tensile or compressive strain on the fiber or a change in temperature will modify the modulation period and hence the reflected wavelength. The fiber-optic sensors have a number of advantages over the electrical alternatives including a high sensitivity. Polymer coating provides good resistance toward water and chemicals, in addition to mechanical protection.

**Table 2. Longitudinal (Vertical) Strain Values at the Proximal Level During Loading of 1000 N. Measured Values are from the External Cortex (Rosette Strain Gauges) and at the Internal Cortex (Fiberoptical Wires) After Implantation of an Uncemented and Cemented Titanium Prosthesis, Respectively. Negative Strain Values Designate Compression**

| | Uncemented #10 Prosthesis<br>Microstrain | Cemented #9 Prosthesis<br>Microstrain | Change with Cement<br>Microstrain* |
|---|---|---|---|
| **Medial** | | | |
| External cortex | - 570 | - 382 | 188 |
| Internal cortex | 0 | - 55 | - 55 |
| **Anterior** | | | |
| External cortex | - 110 | - 260 | - 150 |
| Internal cortex | 210 | 203 | - 7 |
| **Lateral** | | | |
| External cortex | 27 | 55 | 28 |
| Internal cortex | - 90 | 115 | 205 |
| **Posterior** | | | |
| External cortex | - 87 | 36 | 123 |
| Internal cortex | - 56 | 135 | 191 |

* Correlation coefficient: 0.07.

When the femoral neck is cut for insertion of a prosthesis, the biomechanical function of the bone in the proximal femur is severely affected [6, 7]. The common design rationale for femoral stems is to achieve a close fit of the prosthesis to the bone to restore the strain in the proximal femur and to obtain maximum mechanical stability of the implant to minimize micromotion between the implant and the surrounding bone [9, 10]. The mechanical stability is determined by the frictional forces at the implant-endosteal bone interface, the interlocking between the implant and the bone and the elastic deformation of the bone and the implant at their interface. Alterations of the physiological strain patterns induce dynamic bone remodeling which may cause changes in bone density and geometry, and ultimately an unstable situation at the implant-bone interface will occur [7].

It has been a common opinion that stem size and stiffness are the dominant features controlling bone remodeling after hip replacement [1, 11, 12]. However, when titanium implants are compared to stainless steel with identical design, there are small differences in proximal stress shielding [6,13]. We also found some slight differences in strain patterns between a smaller and a larger stem, but it should be emphasized that the stem difference from #9 to #10 was rather small. In addition cementing of a #9 prosthesis with a cement mantle of only 1 mm does not reflect the clinical situation. However, this was done to show the immediate effect of changing an intimate contact between the bone and the surface of the prosthesis to an interface between bone and cement surrounding the prosthetic stem. Our results are in accordance with the opinion that factors contributing to femoral cortical strains are rather complex, and that size and stiffness are less important than conformity with metal implants which are rather stiff as compared to bone [14-16].

After insertion of the implant, strain patterns are unpredictable [13, 17]. Much work has been done to obtain a design of the stem that maintains physiological strain after insertion of the implant [5, 18-20]. Strain gauge measurements usually have been employed to study the pattern of load transfer from the stem to the femur after insertion of a prosthesis. However, on the femur only external surface strain has been presented and often as longitudinal strain only.

Our aim was to measure the strain at the interface between the implant and the internal cortical bone. The strain measurements in this study showed that the strain patterns were significantly different at the internal and external cortex.

Studies of acute changes in the strain pattern in cadaver femurs after insertion of femoral stems are valuable in the assessment of implant variables. Our findings correspond to the results reported by other authors [5, 18, 19]. However, strain studies on the external cortex have been used uncritically to predict the *in vivo* performance, as the stress at the internal interface between the implant and endosteal bone never has been studied. Furthermore, deformation measurements on the external cortex have been assumed to reflect the simultaneous strain at the internal cortex. However, according to finite element analysis [20], there is a shift in stress from periosteal to endosteal surfaces when bone is put under stress. The results of the present study demonstrate that the external cortical condition does not correspond to the situation at the internal cortex. As the internal cortical surface is of outmost importance for the ingrowth of bone and stability of the prosthesis, we urge that future studies should pay attention to this fact.

## ACKNOWLEDGEMENT

None declared.

## CONFLICT OF INTEREST

The authors of this article state that they do not have any conflicts of interest.

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Received: June 8, 2011 Revised: September 1, 2011 Accepted: September 5, 2011