Load distribution in the healthy and osteoporotic human proximal femur during a fall to the side

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Abstract

Due to remodeling of bone architecture, an optimal structure is created that minimizes bone mass and maximizes strength. In the case of osteoporotic vertebral bodies, however, this process can create over-adaptation, making them vulnerable for non-habitual loads. In a recent study, micro-finite element models of a healthy and an osteoporotic human proximal femur were analyzed for the stance phase of gait. In the present study, tissue stresses and strains were calculated with the same proximal femur micro-finite element models for a simulated fall to the side onto the greater trochanter. Our specific objectives were to determine the contribution of trabecular bone to the strength of the proximal femurs for this non-habitual load. Further, we tested the hypothesis that the trabecular structure of osteoporotic bone is over-adapted to habitual loads. For that purpose, we calculated the load distributions and estimated the apparent yield and ultimate loads from linear analyses. Two different methods were used for this purpose, which resulted in very similar values, all in a realistic range. Distributions of maximal principal strain and effective strain in the entire model suggest that the contributions to bone strength of the trabecular and cortical structures are similar. However, a thick cortical shell is preferred over a dense trabecular core in the femoral neck. When the load applied to the osteoporotic femur was reduced to approximately 61% of the original value, strain distributions were created similar in value to those obtained for the healthy femur. Since a comparable reduction factor was found for habitual load cases, it was concluded that the osteoporotic femur was not ‘over-adapted’.

Keywords: Bone mechanics; Osteoporosis; Finite element analysis; Proximal femur; Bone strength

Introduction

According to ‘Wolff’s Law’ bone morphology and strength are adapted to the normal – habitual – loads of daily life. In addition, we experience non-habitual – or ‘error’ – loads in accidents and physical exercises. This brings us to the apparent contradiction that those who engage in sports, for instance, have higher risks for accidents, but at the same time are better protected against fractures. While those who do not, have lower risks for accidents, but are less protected against their occurrence. This condition was earlier baptized over-adaptation [1], and demonstrated for osteoporotic vertebral bodies in micro-finite element (micro-FE) analyses of stress transfer [2]. It implies that bone microstructure is well adapted to daily loads, but vulnerable for ‘error’ loads. The purpose of the present study was to investigate if it can also be demonstrated for the proximal femur.

Using micro-FE models, van Rietbergen et al. [3] evaluated load transfer in a healthy and an osteoporotic human proximal femur during the stance phase of gait. They concluded that the trabecular strain distributions in healthy and osteoporotic bones were quite different, but if the load acting on the osteoporotic femur was reduced to 59% of the normal load, the strain results were quite similar. This suggests that the osteoporotic bone might have been adapted to lower load levels, but is not ‘over-adapted’. Homminga et al. [2] found that strain distributions in healthy and osteoporotic human vertebral bodies were very similar for physiological loading conditions. However, when non-habitual ‘error’ loads were applied, stresses and strains in the osteoporotic vertebrae were much higher than those in the healthy one, suggesting a considerably reduced strength due to ‘over-adaptation’. For the present work, we asked the question...
whether similar effects of over-adaptation could be found in the proximal femur.

A second question concerns the role of the cortical bone shells around the trabeculae. Obviously, proximal femoral strength depends on both cortical and trabecular contributions. Pistoia et al. [4] investigated stresses and strains in the bone tissue of a human radius, using a micro-FE model in which they created bone loss according to several scenarios. They found that reduction of trabecular bone mass had only a small effect on predicted bone strength, whereas a reduction in cortical bone thickness had a major effect. Hence, cortical bone might play a larger role in load transfer and bone strength than often suggested and should be accounted for in analyses of bone strength. The question posed for this work was to what extend this is also a prominent feature in the proximal femur.

We tested cortical and trabecular stress transfer in the proximal femur of two bones, one normal and one osteoporotic, using micro-FE analysis. In each case two situations were compared: stress transfer due to the hip–joint forces of stance – a ‘normal’ load – and stress transfer due to a fall to the side on the greater trochanter, an ‘error’ load. For that purpose we re-used the results of van Rietbergen et al. [3], representing the case of normal loads, and compared those to the effects of ‘error’ loads due to a fall to the side, using the same micro-FE models. We asked the question (1) is the trabecular bone structure in the osteoporotic femur over-adapted to physiological loads? And (2) what are the roles of cortical versus trabecular bone for femoral strength during non-habitual loading?

Materials and methods

This study utilized two micro-FE meshes that were created for an earlier study [3]. The meshes were created from high-resolution computed tomography (CT) images of the proximal 10 cm of a healthy (T-score: −0.5) and a severely osteoporotic (T-score: −4.0) femur. Both femurs were from female donors and were selected from a group of 80 elderly cadavers from an anatomic dissection course. Bone mineral density (BMD) measurements (DPX-L, DXA scanner, Lunar) indicated significantly different BMD values in neck and trochanter of the healthy (0.917 and 0.976 g/cm²) and the osteoporotic femurs (0.496 and 0.656 g/cm²). The donors had closely matched ages (healthy: 82, osteoporotic: 89 years), body weights (63 and 57 kg), lengths (1.60 and 1.61 m) and femoral-head diameters (45 mm each). Using a μCT scanner (μCT-80, Scanco Medical, Bassersdorf, Switzerland), three-dimensional CT images were created with an isotropic spatial resolution of 80 μm, and subjected to a modest Gaussian filter to reduce noise. The images were then segmented using a global threshold value to create binary image that contained bone tissue only. The same threshold value was used for both bones. The bone voxels were directly converted to equally sized 80-μm brick elements, rendering micro-FE meshes of 97 and 72 million elements and 130 and 100 million nodes for the healthy and osteoporotic femurs, respectively.

In each femur, cortical and trabecular bone tissue were identified based on the number of elements in a fixed neighborhood. Each element was assigned a Poisson’s ratio of 0.3 and an isotropic Young’s modulus that depended on tissue type: 15 GPa for trabecular bone and 22.5 GPa for cortical bone [5,6], respectively.

Triaxial boundary conditions applied represent a fall onto the greater trochanter (Fig. 1), based on experimental studies reported earlier [7–9]. The angle between the femoral shaft and the horizontal was 10° and the femur was internally rotated by 15°. The appropriate orientation of the resultant hip force was estimated for both femurs. An arbitrary hip force of 1.0 kN was distributed over the assumed contact areas between pelvis and femur, with the individual nodal forces directed towards the center of the femoral head, to simulate a friction-less cartilage layer. The distribution of nodal forces was first estimated for a perfect sphere [10,11] and then iteratively adjusted until the resultant hip force was pointing into the desired direction, with an error not larger than 1°. The surface nodes in a 0.5-cm layer perpendicular to the resultant hip force on the greater trochanter were fixed vertically to simulate constraints between the femur and a contact surface. Displacement of the nodes at the distal end of the femoral shaft was only allowed in the vertical direction (Fig. 1).

The linear-elastic micro-FE models were solved using an iterative element-by-element solver (Scanco Medical AG, Bassersdorf, Switzerland). The computations for the healthy femur were sufficiently converged (relative errors in the residual forces and displacements were less than 0.001) [19,20] after 40,750 iterations, requiring 17 gigabytes and 3 weeks of wall-clock time on a SGI Origin 3800 supercomputer. The osteoporotic femur required 37,150 iterations and 13 gigabytes of memory. Tissue stresses and strains were used to compute strain-energy densities (SED), effective strains and maximal principal strains.

The elastic energies were computed from the SED data for the trabecular cores and the cortical shells. The ratio of the elastic energies stored in each type of bone is a measure for the contribution of each type of bone to the total load-carrying capacity. The forces on the element nodes were determined in the middle of the femoral neck, in a plane perpendicular to the neck axis. The nodal forces, second moments of area and the trabecular and cortical areas were determined in the selected plane to assess the contribution of both types of bone on a local level.

The yield load of each femur was estimated based on a bone–tissue maximal principal strain criterion [12]. The maximal principal strain – which can be positive (tensile) and negative (compressive) – was computed in each voxel and subsequently smoothed to decrease the fluctuations at the voxel–mesh boundaries and, therefore, improve local accuracy [13]. The yield strains of cortical bone in tension and compression were taken as 0.73% and −1.12%, respectively [14]. The ratio of the yield strains over the maximal principal strain was considered the safety factor for the applied load of 1.0 kN. As such, multiplying the applied hip force with this safety factor results in a yield load that indicates the onset of local failure.

The ultimate loads were estimated based on a failure criterion for micro-FE models of the human distal radius [15]. It uses the effective strain distribution in the entire model, assuming that bone failure occurs when 2% of the bone is loaded beyond 0.7% strain. In contrast with the aforementioned method to determine the safety factor, scaling of the effective strain can result in tissue strains that exceed the assumed tissue yield strains. The estimated yield and failure loads were compared to those based on femoral neck and trochanteric BMD values [7–9].
In the case that the trabecular structure of the osteoporotic femur is ‘over-adapted’ [1], its failure resistance against transverse loading – as in a fall – would be more reduced than its failure resistance against normal hip–joint loading. We know that the failure resistance against normal hip–joint loading for the osteoporotic femur is 59% of the failure resistance of the healthy femur [3]. Hence, the resistance of the osteoporotic femur against transverse loading was compared to this value, to investigate if this bone was indeed ‘over-adapted’.

Results

In both femurs, the highest strains due to a fall occur in the cortex of the femoral neck (Fig. 2) and are opposite in sign to the strains that normally occur in musculoskeletal functions [3]. The higher strains in the osteoporotic femur are evident (Fig. 3).

![Fig. 2. Maximal principal strain in the healthy (A) and osteoporotic (B) proximal femurs. The posterior halves are shown.](image)

![Fig. 3. Maximal principal strain distributions in the healthy and osteoporotic proximal femurs. A reduced hip force of 613 N applied to the osteoporotic femur resulted in the scaled osteoporotic distribution.](image)

![Fig. 4. Maximal principal strain in the selected plane though the femoral neck in the healthy (A) and osteoporotic (B) femur.](image)
The healthy maximal principal strain distribution was compared to a scaled version of the osteoporotic distribution. An external load reduction of 61.3% resulted in the smallest error between both histograms. This scaled distribution, added to Fig. 3, simulated a reduced load on the osteoporotic femur of 613 N.

Broader histograms of the strain-energy density (SED) distribution were found for the osteoporotic femur. The portion of strain energy stored by the trabecular core in the entire models (67.8% and 70.3% for the healthy and osteoporotic femur, respectively) was roughly the same as the portion of trabecular tissue (70.4% in the healthy femur and 73.6% in the osteoporotic femur).

The bone area in the selected plane through the femoral neck (363.6 mm² in the healthy case and 254.3 mm² in the osteoporotic case) is 30.1% less in the osteoporotic case. Since the amount of cortical bone is similar in both femurs (131.0 mm² in the healthy case and 116.6 mm² in the osteoporotic case), this difference is mainly due to resorption of the trabecular core. As a result, the second moments of area $I_{xx}$ and $I_{yy}$, measures for the resistance to bending, are only 9.5% and 11.8% less in the osteoporotic femur (22,128 and 23,409 mm⁴ for the healthy femur and 20,023 and 20,647 mm⁴ for the osteoporotic femur). Plots of the maximal principal strains (Fig. 4) in the selected plane through the femoral neck and nodal forces perpendicular to this plane (Fig. 5) indicate that the cortical shell is subjected to higher strains and carries a larger portion of the load.

The maximal principal strains in the cortex were compared to the tissue yield strains. A hip force of 5.45 kN was needed for the tissue to reach the yield strain of the healthy femur. Tensile yield was reached first in the inferior region of the neck. Increasing the load on the osteoporotic femur from 1.0 to 3.21 kN resulted in maximal principal strains equal to the tissue compressive yield strain in the superior region of the neck.

Failure loads were estimated based on the effective strain distribution [15]. A failure load of 5.83 kN was needed to load 2% of the tissue in the healthy femur beyond an effective strain of 0.7%. The same criterion resulted in an estimated failure load of 3.29 kN for the osteoporotic femur. The failure load for the osteoporotic femur was also estimated from the scaled osteoporotic distribution (Fig. 6). Although a similar distribution was created when scaled with a factor of 0.605, the failure load after scaling (4.7 kN) was lower than the fracture load of the healthy femur.

Discussion

Our first objective was to quantify the stresses and strains in the bone tissue due to a non-habitual load. By comparing the calculated bone–tissue stress and strain distributions with those found in an earlier study for physiological loading conditions [3], it appears that during a fall to the side the stresses do not only increase, but also change, locally, from tensile to compressive, and vise versa. The portion of bone tissue loaded in
tension is larger than that found for physiological loading conditions, leading to almost symmetric distributions for the tissue maximal principal strains in the osteoporotic femur (Fig. 3). The distributions in Fig. 3 are created from the entire meshes, and cannot be compared directly to those found in the earlier study for physiological loading, where only the femoral head was investigated. However, similar distributions were found in the present study when only the head region was included. The increased portion of bone loaded in tension is disadvantageous, since it is stronger in compression than in tension. In fact, the tissue yield point in the healthy femur is reached first in tension.

When the load on the osteoporotic femur is reduced to approximately 60% of the original value, maximal principal strain and effective strain histograms of the healthy and osteoporotic bones are very similar. Interestingly, this value is about the same as found earlier [3] for physiological loading conditions. Hence, the strength in the abnormal loading direction is affected to a similar extent as in the physiological loading direction. In other words: there is no evidence that this osteoporotic femur is over-adapted to physiological loads. This result may seem to contradict results of earlier studies in which it was found that osteoporotic specimens were over-adapted in the sense that their structure was such that strength was maintained for loads acting in the primary load orientation but compromised for loads acting in other directions [16,17,2]. Three points should be discussed though. First, this phenomenon of over-adaptation will occur only for bones that are osteoporotic due to hormonal or other metabolic changes, whereas the loads acting on them are in the normal range. It will not occur, however, for bones that are osteoporotic due to reduced activity or disuse. In the case of complete disuse, a more-or-less uniform decrease of bone material would be expected, because all bone would have equal chances to be resorbed. Unfortunately, for the specimens investigated here, no records of the donors were available (other than that neither of the donors has a disease that would affect the bone). Hence, it is well possible that the osteoporotic specimen was obtained from a donor that had osteoporosis due to disuse, in which case no over-adaptation would be expected. Second, in this study the definition of over-adaptation applied to the entire proximal femur rather than to a piece of trabecular bone. The stress and strain histograms comprised both trabecular and cortical tissue. Although it is possible that locally trabecular over-adaptation is present in the osteoporotic femur, this apparently does not lead to differences in strength of the bone as a whole when comparing two very different loading modes (physiological versus a fall). Third, although loading conditions applied to simulate physiological loading and a fall loading are completely different, they result in a similar stress/strain distribution for most of the bone, be it that regions that were originally loaded in tension are now loaded in compression and vice-versa (compare Fig. 2 with Fig. 3 in van Rietbergen et al. [3]). Hence, forces applied to the trabecular bone regions have changed in magnitude and sign, but no major changes in force direction are expected. This implies that an ‘error load’ (i.e. a load with an unusual direction) for the bone as a whole not necessarily leads to an ‘error load’ for trabecular bone regions. This also could explain why no evidence for over-adaptation is found in the present study. It is possible that other unusual loading modes of the femur (e.g. torsion) would lead to more pronounced changes in load direction for trabecular bone regions.

A second objective was to determine the contribution of the trabecular core to bone strength and to estimate the apparent yield load of each of the bones. We found that the portion of strain energy absorbed by the trabecular core is roughly the same as the portion of trabecular bone tissue in the models. This could indicate that the trabecular and cortical tissues are of equal importance when it comes to energy absorption during a fall. However, the selected plane through the femoral neck shows that the trabecular core is hardly loaded in the osteoporotic femur. The greater part of the load is carried by the cortical shell, which is even more pronounced in the osteoporotic femur on the superior side.

There are a few issues that warrant discussion. First, the boundary conditions applied might not be exactly the same as the conditions that occur during a fall on the side. Instead, they represent the boundary conditions as applied during commonly accepted mechanical tests for the measurement of bone failure load. As in these tests, however, we did not account for the possible role of muscle forces and only static loads were applied. Second, only the proximal 10 cm of the femurs were analyzed. It was found, however, that the maximal strain values were found in the cortex of the femoral neck, far away from the prescribed boundary conditions. Third, in this study, only linear-elastic FE analyses were performed. Such linear models are reasonable only for loads up to the yield point. It has been stated that, in the case of excised trabecular-bone specimens, geometric nonlinearities should be taken into account, even for loads less than the yield load, since geometric nonlinearities could well precede tissue yielding [18]. In our simulations, however, the highest strains were located in the cortical shell. The yield and ultimate loads were, therefore, largely based on the strain distribution in the cortex, such that a geometric nonlinear model becomes of lesser importance. Nevertheless, for more realistic simulations of bone failure beyond the yield point, nonlinear finite element analyses are required. Presently, however, hardware requirements hinder the (materially and geometrically) nonlinear micro-FE analyses for large models as used in this study. Fourth, the ultimate load was predicted based on the calculated tissue strains using a phenomenological method, optimized for compression experiments on the human distal radius. It uses the effective strain distribution, what makes it a robust method. However, it cannot distinguish between failure in tension and compression. Also, since the effective strain in a predefined portion of the tissue in the entire mesh (2%) should exceed 0.007, the predicted failure load depends on the size of the modeled region as well. Surprisingly, the different methods resulted in predicted yield and ultimate loads that differed not more than 7%. Moreover, it was found that both the predicted yield load and the predicted ultimate load were in a realistic range and corresponded well with reported values based on femoral neck and trochanteric BMD values: 5–10 kN and 1–5 kN for the healthy and osteoporotic proximal femurs, respectively [7–9]. Finally, only two proximal femurs were analyzed in this study. The specimens that were selected had
very different density distributions: a healthy proximal femur (T-score: −0.5) versus a severely osteoporotic proximal femur (T-score: −4.0). In spite of this difference, the same decrease in bone strength was found when the femurs were subjected to physiological loading [3] and to a simulated fall to the side. It is, therefore, likely to find similar results for other bones with smaller differences in bone density.

Based on the results of these and earlier [3] analyses, we conclude that the osteoporotic femur analyzed is not ‘over-adapted’. The cortical bone in the femoral neck is relatively highly loaded in the osteoporotic case. Although osteoporosis mainly affects trabecular bone, it has consequences for the loads in the cortical bone as well, in an indirect way. The fact that these highly loaded cortical bone regions remain intact suggests that the load-adaptive mechanisms are still active in the osteoporotic bone.

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References