Discrimination between cases of hip fracture and controls is improved by hip structural analysis compared to areal bone mineral density.

An ex vivo study of the femoral neck

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Abstract

In vivo bone densitometry is affected by measurement inaccuracies arising from the assumptions made about soft tissue and marrow composition. This study tested the hypothesis that section modulus (SM, a measure of bending resistance) when measured ex vivo, would discriminate cases of hip fracture from controls better than areal bone mineral density (aBMD). The biopsies were from (n = 22, female) subjects that had suffered an intracapsular hip fracture. The control material (n = 24, female) was from post-mortem subjects. Serial peripheral quantitative computed tomography (pQCT) 1-mm thick cross-sectional images of femoral neck previously embedded in methacrylate were obtained with the Densiscan 1000 pQCT densitometer and matched for lateral location. The image voxels were converted to units of bone mass, which were then used to derive the section modulus. The data were used to derive means from which receiver operating characteristic (ROC) curves could be generated. The area under the curves (AUC) showed that discrimination between the fracture cases and controls was better for SM than aBMD [SM: AUC = 0.83 (95% confidence interval: 0.71, 0.96), aBMD: AUC = 0.70 (0.54, 0.85); P = 0.034]. To simulate the forces experienced during a sideways fall, the model's neutral axis was rotated by 210°. The results for section modulus were predictable from those at 0° (r² = 0.97). We conclude that biomechanical analysis of the distribution of bone within the femoral neck may offer a marked improvement in the ability to discriminate patients with an increased risk of intracapsular fracture. Progress towards implementing this form of analysis in clinical densitometry should improve its diagnostic value, but may depend in part on better image resolution and more accurate corrections for the variability between subjects in regional soft tissue composition.

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Introduction

A bone’s fragility may be assessed in terms of its bending resistance and its brittleness (or the inverse of brittleness, toughness). These define the extent to which the structure will absorb energy without failing. It is the current practice to assess hip strength clinically using dual energy absorptiometry (DXA) of the proximal femur. However, there are potential difficulties with the calculated areal bone mineral density (aBMD, g/cm²). There are accuracy errors generated by the mass and composition of soft tissue [36]; no account is taken of regional variations in porosity or cortical bone thickness; and with aBMD, the large effects of distance of bone elements from the femoral neck cross section’s centre of gravity is ignored. The effect of the slow tendency of the femoral neck cross section to expand with ageing [19,31] on the femur’s bending resistance is to increase it if bone mineral content is held constant [16,38]; however, cross-sectional area forms the denominator in the calculation of BMD...
from bone mineral content (BMC) and so if it increases, BMD paradoxically tends to fall.

At the femoral neck site, at least 50% of bone strength resides in the cortical shell [22,23]. The femoral neck’s capacity to resist “bending to failure” is increasingly compromised as cortical porosity increases [7,15,24,26]. The mineral content of cortical bone tissue is inversely proportional to its porosity, for the same bone age and in the absence osteomalacia [12]. However, as previous studies have shown, the distribution of bone mass as well as the amount of bone may contribute significantly to the risk of hip fractures [4,5,10].

In the current investigation, we examined the calculated stiffness of the proximal femoral neck, and particularly its calculated resistance to bending measured at its midsection. We made our measurements ex vivo, in part because the biggest uncontrolled factor in clinical bone densitometry is the effect of variable soft tissue composition [6,8,36,40]. We had previously examined the utility of in vivo hip strength estimates from DXA data in a case–control study; the results were promising but required the inclusion, in a multivariate discrimination, of non-densitometric variables such as age to achieve better separation between cases and controls using hip strength endpoints [9]. The measurement of the proximal femur’s structural geometry was achieved by analysing images generated by pQCT allowing definition of its composition. Beam theory was adapted to model the measurements of strength using calculations of the density-weighted cross-section moment of inertia (DWCSMI) and section modulus (SM). This approach was then compared in a Bayesian analysis with conventional DXA-like aBMD measurements, as calculated by projecting the pQCT images from three into two dimensions. The aim was to determine whether SM could form the basis for developing improved risk detection based on noninvasive bone densitometry in either two or three dimensions [e.g. DXA or non-peripheral quantitative computed tomography (QCT)].

Materials and methods

Subjects

Biopsies (n = 22) were obtained from female subjects who had suffered an intracapsular fracture of the femoral neck due to a fall from standing height or a no greater level of trauma. The subjects’ ages ranged from 63 to 89 years (mean 82, SD ± 8.1, median 79 years) and the biopsies were taken 0–6 days after the fracture at the time of repair with prosthesi10ns. The study was approved by the local ethics committees of Cambridge District, Peterborough District, UK, and Kuopio Finland. Written informed consent to participate (giving permission for their clinical records and discarded bone to be used) was obtained from all subjects. There were 24 post-mortem female controls aged between 62 and 100 years (mean 80 ± 7.2, median 82 years) at the time of death (time since death: 3.3 ± 1.3 days, mean ± SD). Their relatives gave permission for use of post-mortem material for research directly after routine hospital post-mortem examinations to establish the cause of death, according to contemporary Hospital procedures. Exclusions included: (i) a history of known diseases affecting bone locally such as carcinoma or Paget’s disease; (ii) renal failure and other causes of osteosclerosis or osteoporosis; (iii) use of bone-active drugs such as glucocorticoids; and (iv) subjects with disuse osteoporosis admission to hospital more than 14 days before death.

Preparation of the biopsies

The femoral neck biopsy specimens from patients with intracapsular fractures were taken during standard hemi-arthroplasty with minimal thermal and mechanical damage [27]. The biopsy specimens were complete cross sections taken from between the line of fracture and the cut face prepared for the insertion of the prosthesis at the base of the femoral neck. The specimens were all obtained at a similar location. The biopsies were fixed in 80% ethanol and embedded in methyl methacrylate without decalcification [4,5,10]. Subsequent to the embedding, the cut face of the biopsy specimen, adjacent to the trochanter, was trimmed and the analyses were completed from this aspect towards the fracture site (or the femoral head in the control biopsy specimens). The inferior cortex was identified by the thickest cortical region [5] and the anterior aspect was identified according to whether the femur was from the left or right side.

Peripheral quantitative computed tomography

The prepared embedded biopsy specimens were fixed by tape to a jig so that both were central in the scan beam of the peripheral quantitative computed tomography (pQCT) (Densiscan 1000; Scanco Medical Ag, Zurich, Switzerland) [10]. This is a second-generation translational and rotational CT scanner that uses a narrow fan beam and a highly collimated detector system to reduce scattering and increase resolution, (maximum 0.195 mm). Serial slices were 1-mm thick. The radiation source has a maximum voltage of 50 kV and an effective energy of 39.5 keV with the X-rays filtered to a narrow energy band. Quality assurance was performed using the Institute for Biomedical Engineering of ETH (IBT) phantom [28] to ensure machine stability (CV over 1 year: area, 0.044%; linear attenuation coefficient, 0.079%).

Scanco AG software was used to allow the preparation of a user-defined control file, which was used for all the specimens. A scout view was taken of the location where the prepared face of the embedded biopsy was adjacent to the jig in order to place the scan commencement reference line. From this line, tomograms 1 mm thick with a 10-cm scan diameter, were made at 0.5-mm intervals for a total
distance of up to 1 cm in the controls (20 slices) or 0.5 cm in the fracture cases (10 slices). The smaller number of slices taken in the fracture cases was due to the restricted amount of material that was available. The first tomogram, which included the prepared biopsy face and the jig, was not used. This left 354 images from control biopsies and 94 from case biopsies.

**Image analysis**

The pQCT images of each slice were imported as TIFF images into the public domain Scion image program v. Beta 4.02 (this software was based on NIH Image originally developed at the National Institutes for Health, USA). An initial macro was used to invert the scanned images and subtract the background grey level, which was set as the mean plus 2 SD of the grey levels associated with methyl methacrylate and soft tissues. The TIFF image data arrays were then analysed as follows. The grey level data were converted to attenuation coefficients on a voxel by voxel basis. Bone mineral density (BMD) values for each voxel were then calculated using the following standard equation:

\[
\text{BMD} = \rho \times \left( \frac{\mu(gv) - 0.241}{1.423 - 0.241} \right)
\]

where \( \rho \), the density of mineralised bone tissue without any reduction for the effect of porosity (in g/cm\(^3\)) took the generally accepted value 2.2; the attenuation coefficient of the individual voxel element’s grey level value is \( \mu(gv) \); the attenuation coefficient of bone marrow took the value of 0.241 cm\(^{-1}\); and the attenuation coefficient of bone tissue took the value of 1.423 cm\(^{+}\) [18]. As a result of these calculations, each voxel was able to generate a BMD value between the density of bone marrow and the density of bone tissue itself (2.2 g/cm\(^3\)). Cortical voxels with density values below 2.2 were assumed to include a proportion of osteoid, undermineralised bone or the contents of cortical canals. Voxels in the cancellous region were assumed to be composed of variable proportions of bony trabeculae and bone marrow.

Each cross-sectional image array was examined (using MATLAB Version 6.5.0.18091a Release 13) and its size and density calculated. Each voxel element in an image array represented 0.195 × 0.195 × 1.0 mm depth; the maximum height (the \( z \)-axis) and the maximum width (the \( x \)-axis) of the mid-neck cross section were determined from the product of the maximum number of elements for each axis and 0.195 mm. The density of each element was calculated as the product of the volume (0.195 mm\(^2\) × 1 mm) and the BMD calculated from its individual grey level.

**Biomechanical parameters**

The principles of beam theory were applied to measure the theoretical capacity of the midsection of the femoral neck to resist bending. This was done by calculating according to the standard beam equation [38]:

- the location of the neutral axis of the cross section;
- an index of the resistance to pure bending, that is, the density-weighted cross-section moment of inertia (DWCSMI) with reference to a defined loading plane (see below);
- the section modulus as DWCSMI divided by the maximum distance to bone surface in the loading plane [20,39].

The initial calculations of DWCSMI and SM were done with the neutral axis at 0° (horizontal), that is if the loading was in upright stance. The load, amounting to approximately 68% of the subject’s weight [42], would therefore be perpendicular to this neutral axis, with the inferior cortex under compression and the superior cortex under tension [39]. The second application of DWCSMI and SM calculations was done with a neutral axis rotated 210° (anti-clockwise looking towards the femoral head), that is if the subject had fallen on to the greater trochanter. This imposes the greatest compressive and tensile stresses on the superior–posterior and inferior–anterior aspects of the femoral neck, respectively [22].

**Damaged specimens**

Six of the potentially eligible fracture case biopsies had small to moderate portions of cortical shell missing. This may have been due to damage caused when the specimens were taken or more likely fragmentation during the moment of fracture. To exclude these cases would have risked introducing bias (because they might have been the most fragile), yet to include cases with more than a small proportion of their cortex missing would have been to introduce unacceptable conjecture. Before assessing the incomplete biopsies, we arbitrarily defined a maximum acceptable percentage of missing cortical bone at 10% of the total cross-sectional area of cortical bone. In the six cases that were amended, the missing proportion of bone was less than 10% bone mass and area (range 1–9%).

We then “restored” the cortical cross section by “painting in” the missing elements on the spreadsheet recording the voxel density data. This was done as follows: the mean level of BMD of the missing voxels was estimated to be the same as sample groups of cortical pixels adjacent to the damage site calculated from the biopsy’s spreadsheet. The outline of the missing cortex was then traced in by extrapolating the contours of the subperiosteal and endocortical margins of the cortex from either side of the defect. Within the newly defined area of missing cortex, each voxel was assigned a BMD value equal to the mean BMD in the adjacent cortex calculated as described above (see Fig. 1).
Estimate of areal BMD

It was assumed that differences in cortical density were more likely to be the result of differences in porosity rather than mineralisation. Cortical areas were therefore adjusted by multiplying the ratio of measured cortical density to the assumed density value of fully mineralised bone. To estimate the areal BMD that would be generated by a DXA scanner, the bone mass of the pQCT cross sections were used and calculated as described earlier by Crabtree et al. [10].

Statistical analysis

A previous study showed that where the ratio of the distances between the superior to inferior (maximum) and anterior to posterior diameters (minimum) is 1.4, this region of the femoral neck has the lowest cross-sectional area and...
can be considered the ‘mid-neck’[10,21]. To determine the relative position of each tomogram along the femoral neck, a linear regression model was constructed which analysed the ratio of maximum/minimum diameter of the individual slices as the dependent variable (JMP v 4.0.2; SAS Institute, Cary, NC, USA) and the distance of each slice from the biopsy methacrylate face, and any interaction between the cases and controls were used as the independent variables.

Linear mixed modelling (SPSS Inc.© 1989–2002 Release 11.5) was used to fit a repeated-measures analysis of variance model to predict section modulus and BMD at six sections taken at intervals of 0.5 mm along the femoral neck towards the femoral head. The origin was located at the mid-neck cross-section where max/min = 1.4 and the most proximal section was at 2.5 mm from the origin. Data from more proximal regions were sparse in the cases because of the fractures that had occurred and so these more proximal data were excluded. Case/control status was included in the model as a between-specimen factor to test for differences between cases and controls and age was adjusted for as a covariate. An interaction between case/control status with distance was also fitted in the model to test if differences between cases and controls were similar in all six sections. Finally, random effects for intercept and slope were defined to model the random variation in the intercepts and slopes for individual specimens.

To examine the possible effects of fracture status on relationships between outcome variables correlated, respectively, in the stance and fall configuration, covariance analysis was used.

Receiver operating characteristic (ROC) curves were generated at each neck level and compared for the purposes of assessing the ability of two different measures of bone properties to discriminate between cases and controls. The areas subtended by the two ROC curves at each level were compared using the ROCCOMP function in the STATA software [13].

Finally, to simulate the accuracy errors generated by soft tissue in vivo, the effects of adding random noise to the extent determined by Svendsen et al. [36] was explored in silico by using a Monte Carlo simulation technique. Thirty sets of simulated data were created that included random noise attributable to soft tissue variability. The noise was created by a computer-driven random noise generator giving a mean of zero and a coefficient of variation of 6.5% for the soft tissue error component (as appropriate for the mid-femoral neck [36]) and added to the appropriate data from the original data set. Section modulus and BMD were calculated for each subject in each simulated data set. Areas under the ROC curves for each of the 30 simulated data sets were calculated and the means and standard deviations of these simulated sets of ROC curve subtended areas were calculated. The areas under the ROC curves for Monte Carlo simulation were compared for aBMD and section modulus using paired t tests.

Fig. 2. Receiver operating characteristic (ROC) for the discrimination of cases from controls for either areal BMD or section modulus.
Results

Comparison of complete and incomplete biopsy cross sections

The six fracture case images that required amendment to “paint in” up to 9% of their cortical cross section were compared with the 16 fracture cases with complete cortical shells. Although the incomplete shell cases had lower SM and DWCSMI as anticipated, there was no significant difference between the two groups (amended cases: SM 422.5 ± 74.3 mm³, mean ± SEM).

Loading in the normal stance position

At the mid-femoral neck, the results for SM showed a significant difference between the fracture cases and the controls (Fig. 3) [cases: 535.2 ± 46.2 mm³ (mean ± SEM) n = 20; controls: 776.9 ± 42.4 mm³, n = 24, P = 0.0001]. The difference between the means of the fracture case areal BMD compared to the controls was somewhat less significant (cases: 0.514 ± 0.02 g/cm²; controls: 0.598 ± 0.02 g/cm², P = 0.008; see (Fig. 3b).

The femoral neck subjected to loads when in a simulated fall configuration

In the fall configuration, the SM results for the mid-femoral neck were lower, as expected, than in stance (cases: 434.02 ± 37 mm³; controls: 617.6 ± 34 mm³; P = 0.001). A Pearson correlation of SM for the stance and fall configurations, where both cases and controls were grouped, showed that the SM values were closely related (r² = 0.976). Covariance analysis showed no significant difference between cases and controls in this relationship (P = 0.8).

Fig. 3. Comparison of cases and controls with regard to: (a) section modulus at mid-femoral neck position (0 mm) (cases 535.25 ± 46 mm³, controls 776 ± 42 mm³, P < 0.0001); and (b) areal BMD at mid-femoral neck position (0 mm) (cases 0.513 ± 0.02 g/cm², controls 0.597 ± 0.02 g/cm², P < 0.008). The data are presented as the mean ± SEM.
Bayesian analysis of section modulus and areal BMD as discriminators of hip fracture cases from controls

ROC curves were constructed for section modulus (stance configuration) and aBMD (see Fig. 2). When the areas under the two ROC curves were compared, there was a better chance of correct discrimination of cases from controls with SM compared to aBMD at all six sections examined from the mid-neck level proximally. At mid-neck, the significance of this difference was $P = 0.034$ and the difference was significant ($P < 0.05$) at the next three sections. However, the difference at the proximal two sections was less significant ($P > 0.1$).

When the differences between cases and controls were plotted as a function of distance from mid-neck, it could be seen that SM decreases proximally in the controls ($54 \pm 7.8 \text{ mm}^3$, slope and SEM) but not in the cases ($1.6 \pm 11 \text{ mm}^3$), whereas aBMD was unrelated to distance from mid-neck in both cases and controls (Fig. 3).

Simulated effect of variability of soft tissue density

When random noise was added to the ROC curve data to simulate the effect of variability of soft tissue composition, the mean area under the ROC for section modulus was 0.78 (as expected from Fig. 3) with a standard deviation of 0.007. For areal BMD, the mean area under the curve was 0.64 and the standard deviation was 0.013.

Discussion

The purpose of this study was to compare the bending resistance of femoral neck cross sections from intracapsular fracture cases with those of controls to see if calculated bending resistance better discriminated cases from controls than areal BMD. This ex vivo study supports the suggestion that the mechanical properties of the femoral neck are considerably less good in cases of hip fracture than in controls, and that with improvements in technology, a useful improvement in hip fracture prediction would be possible using clinical measurement of the femoral neck. This would require more accurate measurements of the proximal femur’s mechanical properties, particularly the distribution of bone mineral about the femoral neck axis. This could in principle be obtained noninvasively from the living bone with DXA or QCT scanning.

Our data also suggest that the main impediment to improving hip fracture prediction from three-dimensional QCT measurements is limited resolution (in two or three dimensions), since the effects of variability in soft tissue composition were insufficient to mask the better discriminating performance of SM in this ex vivo study.

The calculated results for aBMD are what would be expected from a DXA scan if the soft tissue component was subtracted accurately [25]. In practice, the pixel size is different (DXA and Siemens Somatom 16 CT: 0.5 mm by 0.5 mm; pQCT: 0.195 mm by 0.195 mm). In addition, the accurate identification of the amount of bone mineral content is dependent on the position of the femoral neck in relation to the DXA scan table. The femoral neck must be parallel to the scan table and it is well known that incorrect angulation creates inaccuracies. This difficulty maybe overcome by CT as the results produce a three-dimensional image that enables the correct angle to be obtained.

The results of modelling the data in the “fall” projection (-210° of rotation compared to the AP or stance projection) did not change the discrimination between the cases and controls. Moreover, the results for both SM and aBMD, while numerically different, were highly predictable from the results in the AP projection. This supports the notion that clinically valid analyses in two-dimensional of SM and other bone strength parameters may be made from conventional DXA scans.

Properties that might affect the bending resistance of the femoral neck include: its shape, structure, composition and the distribution of the material of which it is composed. The significant decline in the controls’ SM shown in the results (Fig. 3) is due to shape change combined with a decline in the bone mass. In contrast, the lack of change in SM in the fracture cases is due to the bone mass at the mid-neck position remaining similar at each slice moving towards the femoral head. There may however be a change in the distribution and location of the bone mass that has not been identified and should be a subject for further investigation. The properties of this material are also critically important. We could not examine without mechanical testing the possibility that femoral neck bone is more or less tough (i.e. less or more brittle) than control bone. However, when loading of the femoral neck is severe, as during a fall, the elastic limit of the bone tissue is more likely to be exceeded and in consequence, plastic deformation will take place when bending resistance is reduced. Depending on its material toughness, a bone will suffer complete fracture under a greater or lesser stress increase once plastic deformation begins. Thus, the femoral fracture patients we studied could be shown to be at a considerable mechanical disadvantage in the event of loading through a fall onto the greater trochanter. Their bone tissue would have to be considerably tougher for them to have maintained a similar yield strength to that of the controls given an equivalent fall-type loading. As with any composite material, an increase in porosity alters the mechanical properties of the material. Resistance to bending of bone tissue appears to decrease exponentially with increasing porosity [12]. To study the effects of porosity and other factors, regional differences in the distribution and the structure of the cortical bone in the femoral neck have been measured using histomorphometry. Cortical bone porosity was increased in the anterior region of the mid-cross-section of the femoral neck from cases of intracapsular hip fracture compared with age-matched cadaveric material [4].
As discussed by Crabtree et al. [9,10], cortical bone strength is dependent on the degree of mineralisation and porosity [12,34], and the location around the axis through which it is loaded [3]. A major change that affects the bending resistance of any bone depends on its tendency to expand with age [35]. Endosteal resorption tends normally to be balanced at least in part by subperiosteal apposition during growth, a process which slows considerably but does not stop completely [1] in adult life, even in the femoral neck [1,19,31]. In circumstances where the amount of bone tissue remains constant, so that a tubular bone grows in diameter, there will be an increasing resistance to bending although the cortex compensates for this expansion by becoming somewhat thinner. This is explained by the engineering principle (first described qualitatively by Galileo Galilei [16] and quantitatively by Euler [38]), in that the further away mass is distributed from its central, or neutral bending axis, the greater the resistance to bending [2].

When the endosteal rate of resorption is greater than the periosteal rate of apposition, there will be thinning of the cortical bone [30]. A critical situation might occur when this cortical thinning is greater in one region of the femoral neck than in others [3,5,30]. Bell and Power et al. [3,5,30] observed an excess of cortical thinning which occurred in the inferoanterior and superoposterior cortices. This, it was hypothesised, would make these two regions are particularly vulnerable when subjected to unusual strain, such as when the trochanter has impact with the ground during a sideways fall [22,29]. However, in the present investigation, the analysis of SM in the fall configuration showed that although it was reduced by 20%, there were similar relative differences between cases and controls to those found in the stance configuration.

To calculate the SM measurements that would be appropriate when the subject was experiencing a fall sideways, the neutral bending axis has to be rotated by more than 180° [22]. The load on the proximal femur at the time of impact after a sideways fall results in the inferoanterior cortex being subject to tensile stress and the superoposterior cortex being subject to compressive stress; these stresses represent a reversal of the normal bending in stance and it has been suggested that the strength, derived from the orientation of a bone’s anisotropy and geometry, will be reduced in this circumstance [14,17,32,33,37].

The diagnosis and assessment of osteoporosis with planar DXA densitometry is dependant on the value of areal bone mineral density, that is, bone mineral content divided by area. This typically produces mean case–control differences of 14% [11] as has been reproduced in this study. The differences in discrimination, between the areal BMD and measurements of SM, emphasises the importance of quantifying the composition and distribution of materials when analysing an inhomogeneous compound structure. The accurate identification of the bone’s component materials that provide strength, and the distribution of porosity, is critical in assessing accurately an individual bone’s ability to resist bending.

The biomechanic analysis presented here clearly provided better discrimination than aBMD when these results were subjected to a Bayesian analysis using receiver operating characteristic (ROC) curves (Fig. 2). If the clinical assessment to therapy were to be simulated from this data, if specificity was set at 80% and the 20% of subjects at most apparent risk were selected for intervention using SM then 76% of the subjects would stand to benefit. If areal BMD was used similarly, then only 38% of the subjects would benefit at the same level of specificity (Fig. 2). We have however obtained in this ex vivo high-resolution pQCT study greater enhancement of SM over aBMD than seen in our earlier in vivo DXA study [9]. This highlights the potential for technical improvement in quantitative analysis of hip densitometric images in support of accurate measurements of hip SM.

We have shown that the effect of variable soft tissue composition is not likely to be a major confounding factor in vivo. The reduced image resolution of most whole body QCT (and in two dimensions of most DXA) equipment compared to our pQCT Densiscan image might however seriously impair discrimination, since SM and CSMI depend on precise localisation of bone tissue with respect to the neutral axis of the femoral neck. There is also the effect of overestimation of the areal BMD generated by adipose tissue when using DXA in vivo to consider [6,8,36,40,41]. The challenge now is to determine to what extent these problems in the physics of clinical imaging of the proximal femur can be overcome sufficiently to allow CT or DXA studies of the proximal femur to provide the enhanced discrimination using SM that we have shown to be possible ex vivo.

This study has some limitations. The principal limitation is that the moderate number of fracture cases and controls combined with the inability of case–control studies to eliminate undetermined bias means that the quantitative estimates of case–control differences and areas under ROC curves we have presented must be treated with caution. It would be valuable to repeat this study in a prospective framework, but this would only be possible using in vivo imaging for obvious reasons. However, the effect of selection bias on the relative abilities of the aBMD and SM to discriminate cases and controls is likely to be small because the comparison was made between the same cases and controls for the two types of measurement. In addition, the fracture case series of pQCT tomograms had a limited number of scans where the cortical shell was incomplete. When those cases were excluded which had up to 9% of their cortex amended because of damage, the differences between areas under the ROC curves actually increased from 0.21 to 0.27. So including these cases did not confound our results.

In conclusion, while the mean areal BMD of the fracture cases was only about 10% lower than that in the controls,
the difference in the mean SM of the fracture cases and controls was nearly 3-fold greater. With the application of stress analysis techniques, measures of hip strength should provide a substantial improvement in the ability of diagnostic imaging to discriminate between hip fracture cases and controls. The rotation of the femoral neck to align the scanning plane, so as to measure SM in the fall-loading configuration, did not affect discrimination. In practice, the improvement in discrimination from the use of hip strength analysis will be affected by the resolution of scanning systems; thus, a DXA system with a resolution of considerably greater than 0.2 mm could not be expected to show the level of discrimination in vivo that is possible with our ex vivo pQCT study. The improvement of both bone and soft tissue image resolution and their accurate localisation are therefore important future targets in bone densitometry of the hip.

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References


