Published in final edited form as:

J Bone Miner Res. 2009 January ; 24(1): 33–42. doi:10.1359/jbmr.080906.

Use of DXA-Based Structural Engineering Models of the Proximal Femur to Discriminate Hip Fracture

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Abstract

Several DXA-based structural engineering models (SEMs) of the proximal femur have been developed to estimate stress caused by sideways falls. Their usefulness in discriminating hip fracture has not yet been established and we therefore evaluated these models. The hip DXA scans of 51 postmenopausal women with hip fracture (30 femoral neck, 17 trochanteric, and 4 unspecified) and 153 age-, height-, and weight-matched controls were reanalyzed using a special version of Hologic's software that produced a pixel-by-pixel BMD map. For each map, a curved-beam, a curved composite-beam, and a finite element model were generated to calculate stress within the bone when falling sideways. An index of fracture risk (IFR) was defined over the femoral neck, trochanter, and total hip as the stress divided by the yield stress at each pixel and averaged over the regions of interest. Hip structure analysis (HSA) was also performed using Hologic APEX analysis software. Hip BMD and almost all parameters derived from HSA and SEM were discriminators of hip fracture on their own because their ORs were significantly >1. Because of the high correlation of total hip BMD to HSA and SEM-derived parameters, only the bone width discriminated hip fracture independently from total hip BMD. Judged by the area under the receiver operating characteristics curve, the trochanteric IFR derived from the finite element model was significantly better than total hip BMD alone and similar to the total hip BMD plus bone width in discriminating all hip fracture and femoral neck fracture. No index was better than total hip BMD for discriminating trochanteric fractures. In conclusion, the finite element model has the potential to replace hip BMD in discriminating hip fractures.

Keywords

osteoporosis; hip fracture prediction; structural engineering model; DXA; biomechanics

INTRODUCTION

HIP FRACTURE IS the most serious complication of osteoporosis, with a high rate of morbidity and mortality in the elderly. Preventative interventions targeted to high-risk individuals can reduce the incidence for hip fractures, and it is therefore very important to identify fracture-prone individuals.

The current assessment of osteoporosis and fracture risk is based primarily on BMD as measured by DXA. Although total hip (TH) BMD has been shown to be the main risk factor for hip fracture, substantial overlap exists in BMD between people with and without hip fracture.^(1,2) Therefore, there is a need to develop more accurate tools of assessing fracture risk using not only BMD but also other predictive parameters.

Hip structure analysis (HSA) uses information about the bone geometry and mass distribution embedded in DXA hip scans to produce parameters that are related to bone strength. The parameters include hip axis length, neck-shaft angle, cross-section bone area, bone width, moment of inertia, section modulus, cortical thickness, and buckling ratio. These have been studied as predictors of hip fracture.^(3–20) It has also been shown that predictors for femoral neck (FN) and trochanteric (TR) fractures of the proximal femur were different.^(6,15,18,21)

A bone fractures when stress within the bone exceeds its ultimate strength. The stress depends on the geometry, spatial distribution, and mechanical properties of the bone, as well as the direction, magnitude, and position of the force applied. Structural engineering models (SEMs) integrate density and geometric information embedded in DXA scans with the applied force. This enables the evaluation of stress throughout the bone and therefore is of potential value in predicting hip fractures. Indeed, several SEMs have been developed that show superior accuracy in the determination of the bone strength over BMD alone.

Based on a technique developed by Martin and Burr,⁽²²⁾ Beck et al.⁽²³⁾ developed and verified a technique to measure the cross-sectional mechanical parameters using DXA. Using this technique, they⁽²⁴⁾ developed a curved-beam model of the proximal femur. In a cadaver study,⁽²⁵⁾ they found that the calculated failure load based on this model was closely correlated to the measured failure load ($r = 0.91$). However, we are not aware of any clinical studies that use stresses derived from this model to predict hip fracture. Applying a similar technique to clinical trial data, Crabtree et al.⁽¹⁴⁾ showed that age, body mass index, and compressive stress at the narrowest FN as a result of a typical fall on the greater trochanter was significantly better at predicting hip fracture than FN BMD alone, and Faulkner et al.⁽¹⁹⁾ reported that FN BMD, hip axis length, and femoral strength index at cross-section with minimum moment of inertia were significant independent predictors of hip fracture. Testi et al.⁽²⁶⁾ developed a finite element (FE) model of the proximal femur based on DXA. In a clinical study involving 47 hip fracture and 52 control patients,⁽²⁷⁾ they showed that including BMD, height, neck-shaft angle, and maximum tensile strain from this model into the regression analysis enhanced the prediction accuracy from 64.5% for BMD alone to 81.7%. Recently, we developed a curved composite beam model of the proximal femur that produced a similar stress distribution to FE models.⁽²⁸⁾

The aim of this retrospective, case-controlled study was to determine whether the curved-beam (CB), curved-composite-beam (CCB), and FE models of the proximal femur have a role in the identification of postmenopausal women at risk for hip fracture, either as a single condition or separately for FN and trochanteric fractures. Based on the fact that the SEMs integrate density, geometry, and loading condition, we hypothesize that the SEM-derived parameters are superior in discriminating hip fracture over BMD alone.

MATERIALS AND METHODS

Case and control populations

We retrospectively studied two groups of postmenopausal women. One group comprised 51 women (age, 58–81 yr) who had sustained a recent low-trauma hip fracture: 30 of the fractures had occurred at the FN, 17 were classified as trochanteric, and 4 fracture sites were

not specified. The classification of hip fracture was confirmed by radiograph and orthopedic notes by a single reader. DXA hip scans were performed on the nonfractured side within 8 wk after fracture using QDR 4500 (Hologic, Bedford, MA, USA). Patients with bilateral hip fractures or those unable to give informed consent were excluded.

For each case, three age-, height-, and weight-matched controls were randomly selected from the population-based OPUS (Osteoporosis and Ultrasound) study,⁽²⁹⁾ which recruited a cohort of 500 postmenopausal women (age, 55–79 yr) from general practice lists in Sheffield. DXA hip scans were performed on the left hip using the Hologic QDR 4500 following an identical standardized protocol.

Development of SEMs

A pixel-by-pixel BMD map of the hip was obtained for each subject by reanalyzing the DXA hip scans using special analysis software provided by Hologic. A suite of Matlab (The Mathworks, Natick, MA, USA) programs was developed to analyze the BMD maps. The proximal femur was segmented semiautomatically, and region of interest (ROI) of various sorts was defined (Fig. 1). The FN and shaft axes were determined automatically followed by the FN, TR, and shaft ROIs. The FN starts from the hip joint center and ends at the intertrochanteric line (Fig. 1, k-e-n). The trochanter ROI starts from the intertrochanteric line and ends at the point on the shaft axis that has the same distance to the neck–shaft axis intersection as the end of FN ROI (Fig. 1, u-f-v). The axis of the trochanter ROI was defined as a circular arch that is tangential to the FN and shaft axes at each end, respectively (Fig. 1, e-m-f). Cross-sections perpendicular to the axis of FN, trochanter, and shaft ROI were defined.

We generated a curved-beam (CB) model, a curved-composite-beam (CCB) model, and an FE model of the proximal femur from each BMD map. The models simulated a typical lateral fall on the greater trochanter. The impact force was perpendicular to the shaft axis and at the level of the intersection point of the FN–shaft axes (Fig. 2). The force magnitude was determined based on the subject's weight and height according to van den Kroonenberg et al.⁽³⁰⁾ From the SEMs, the stress perpendicular to a cross-section (normal stress) at each pixel of a cross-section was calculated. Figure 3 shows the typical distributions of BMD and normal stress derived from the CB, CCB, and FE models. The details of the models are presented in the Appendix. We divided the normal stress at each pixel by the yield stress of that pixel and defined an index of fracture risk (IFR) for the FN, TR, and TH as the mean stress ratio in those regions. The yield stress is depended on volumetric BMD and was determined according to Keyak's equation.⁽³¹⁾ We evaluated IFR as hip fracture discriminators.

HSA

We reanalyzed all the 204 DXA hip scans using Hologic APEX version 2 software that has a HSA option.⁽³²⁾ The cross-sectional bone area, bone width, section modulus, cortical thickness, and buckling ratio averaged over the narrow FN and trochanter ROIs were analyzed in this study.

Statistical analysis

All parameters were tested for normality, and nonparametric and parametric tests were used where appropriate. The OR of fracture for a 1 SD change in parameter value was derived from the univariate conditional logistic regression for matched case-control groups. The ORs adjusted for TH BMD were also calculated to determine whether a parameter was an independent discriminator. We chose TH BMD because we found it a good discriminator for all hip fracture types. To identify the best regression model for each type of hip fracture, we

performed multivariate conditional logistic regression, including TH BMD and the independent discriminators as covariates. The backward elimination method was used to exclude the independent discriminators from the regression model according to likelihood ratio. The best model was compared with TH BMD alone and the best SEM-derived model alone. The comparison was based on the area under the receiver operating characteristics (ROC) curve as well as the sensitivity at 80% specificity and specificity at 80% sensitivity. Cross-validation of the regression models was performed using the bootstrap method. Conditional logistic regression was performed on a randomly chosen 80% of cases and their matched controls. Two cut-off points in probability that corresponded to 80% specificity and 80% sensitivity were chosen and used to classify the remaining 20% subjects and to calculate sensitivity and specificity. This process was repeated 500 times, and the mean sensitivity and specificity in classifying the remaining 20% subjects were calculated and compared. Statistical analyses were performed using Stata 9 (StataCorp).

The analysis was performed regarding the hip fracture as a single condition and repeated for femoral neck and trochanteric fracture cases and their controls.

Validation and precision of the SEMs

To verify the Matlab programs implementing the CB and CCB models, a DXA scan of an idealized proximal femur was generated and analyzed, and the results were compared with the theoretical results (details in Appendix). The maximum error was <5%. To assess the reproducibility of the analysis procedures, 27 scans were randomly selected from this study and analyzed twice, 1 wk apart. The maximum CV of all HSA and SEM parameters was <5%.

RESULTS

All hip fractures versus controls

Table 1 lists the OR (95% CI) of increased hip fracture risk associated with 1 SD change in parameter values. Lower BMD, smaller cross-section bone area, wider bone, smaller section modulus, thinner cortex, higher buckling ratio, and higher IFR were all associated with a significant increase in hip fracture risk, and they were significant discriminators of hip fracture on their own. It was noticed that the OR of the FE-derived trochanteric IFR was similar to that of TH BMD.

After adjusting for TH BMD, only the ORs associated with an increase in bone width remained significantly >1 (width in the narrow neck: OR, 2.1; 95% CI, 1.3–3.5; width in the narrow trochanter: OR, 2.3; 95% CI, 1.3–4.1). Other parameters were not independent from TH BMD as hip fracture discriminators and they were found to be highly correlated to TH BMD (Table 2). Multivariate conditional logistic regression using TH BMD and backward selection of the two independent HSA predictors identified a model that included TH BMD and narrow neck width.

Three regression models, using TH BMD alone, TH BMD plus narrow neck width, and FE-derived trochanteric IFR alone as covariate(s), respectively, were compared in terms of area under the ROC curves (AUC), and the results are shown in Fig. 4A. Adding the narrow neck width to the TH hip BMD resulted in significantly ($p < 0.05$) larger AUC (0.832; 95% CI, 0.774–0.890) than that of TH BMD alone (0.771; 95% CI, 0.705–0.837). The AUC of the model with FE-derived trochanteric IFR (AUC, 0.817; 95% CI, 0.758–0.877) was significantly larger than that of TH BMD alone but slightly, not significantly ($p < 0.05$), smaller than that of TH BMD plus the narrow neck width.

Table 3 shows the results of bootstrap cross-validation of the regression models in term of sensitivity at 80% specificity and specificity at 80% sensitivity. For comparison, the original classification accuracy of the models was also read off from the ROC curves and included in Table 3. Both sets of results were almost identical. The two models, one with FE-derived trochanteric IFR and other with TH BMD plus narrow neck width, had much higher sensitivity and specificity than the model with TH BMD alone, whereas there were small differences between them. The results also indicated that the models did not suffer from the problems of overfitting.

In summary, all the BMDs and SEM-derived parameters and almost all the HSA parameters were significant hip discriminators on their alone. Because of the high correlations of TH BMD to HSA and SEM-derived parameters, most of HSA- and all SEM-derived parameters did not discriminate hip fracture independently from TH BMD. FE-derived trochanteric IFR was significantly better than TH BMD alone and similar to TH BMD plus narrow neck width in terms of AUC and discrimination accuracy.

FN fractures versus controls

Similar to the results comparing all hip fractures versus controls, a lower hip BMD, longer hip axis length, smaller cross-section bone area, wider bone, thinner cortical thickness, higher buckling ratio, and higher IFR were associated with a significant increase in FN fracture risk: ORs for these parameters were significantly >1 (Table 1). Again, only the bone widths discriminated FN fracture independently from TH BMD (narrow neck width: TH BMD-adjusted OR, 2.1; 95% CI, 1.1–4.1; narrow trochanter width: TH BMD-adjusted OR, 2.4; 95% CI, 1.1–5.2). Combining TH BMD and narrow neck bone width was identified in the multivariate regression using TH BMD and backward selection of the two independent discriminators. It was noticed that a regression model using FE-derived trochanteric IFR alone was associated with a significantly ($p < 0.05$) greater AUC and higher discrimination accuracy compared with TH BMD alone (Fig. 4; Table 3). It even had a larger AUC than the model with TH BMD plus narrow neck width. In cross-validation, there was a moderate reduction in the sensitivity compared with the original model, indicating a possibility of minor overfitting.

Trochanteric fractures versus controls

Unlike in FN fracture versus control, the TR fracture did not associate significantly with lower FN BMD, longer hip axis length (HAL), smaller narrow neck CSA, and higher SEM-derived IFR in the FN (Table 1). Similar to the FN fracture cases versus controls, lower BMD, narrower bone, thinner cortex, larger buckling ratio, and higher IFR in the trochanter and total hip were significant TR discriminators on their own. After adjusting for TH BMD, no HSA- and SEM-derived parameters were found to be the independent discriminators of TR fracture. Although the AUC associated with the FE-derived trochanteric IFR (AUC, 0.821; 95% CI, 0.717–0.925) was slightly higher than that of TH BMD alone (AUC, 0.818; 95% CI, 0.717–0.919), the difference was not significant.

DISCUSSION

The purpose of this study was to determine whether DXA-based SEMs have a role to play in discriminating between patients with hip fracture and control subjects. We found that SEM-derived IFRs on their own significantly discriminated hip fracture either as a homogeneous condition or separately as FN and TR fractures. Furthermore, FE-derived trochanteric IFRs alone discriminated all hip and FN fractures significantly better than TH BMD alone and similar to TH BMD plus an independent geometrical parameter (narrow neck bone width).

This supports our hypothesis that the FE model is superior than TH BMD alone because it integrates density and geometry information embedded in DXA scans.

Our results are in agreement with other similar studies. Faulkner et al.⁽¹⁹⁾ reported that femoral strength index, defined as the ratio of compressive stress to yield stress at the FN cross-section of minimum moment of inertia, was significantly higher in hip fracture. Crabtree et al.⁽¹⁴⁾ showed that the compressive stress at the narrowest FN cross-section (Ctress) was significantly higher in the fractured cases than the control and that Ctress combined with age and BMI provided significantly better prediction of fracture than FN BMD alone. Testi et al.⁽²⁷⁾ reported that including a FE-derived parameter, maximum tensile strain, in a regression analysis enhanced the prediction accuracy of hip fracture from 64.5% (BMD alone) to 81.7%. Although we derived the IFR from ROIs rather than from a single cross-section, our results also showed that strength parameters are significant discriminators of hip fracture and superior over BMD.

We found that BMD and SEM-derived parameters did not independently predict hip fractures because of the high correlation between these measurements. The high correlation between BMD and HSA and stress parameters have also been reported by Ahlborg et al.⁽³³⁾ Our data suggest that the FE-derived parameter has the potential to replace BMD as a hip fracture discriminator. Much larger studies are needed to quantify the improvement that might follow the replacement of BMD with FE-derived IFR.

Three different SEMs were used in this study, and they seemed to have different abilities to predict hip fractures. As judged by the OR and AUC, the FE and CCB models performed much better than the CB model, and the FE model was marginally superior over the CCB model. These differences may reflect the inherent differences in the models themselves. One major difference is the distribution of normal stress. The CB model assumes a single material, and the differences in BMD were accounted for in the calculation of the pixel thickness (proportional to BMD). For a particular cross-section, the moment of inertia is a constant. The normal stress caused by bending moment is proportional to how far the point under consideration is away from the cross-sectional centroid divided by the moment of inertia, independent of the BMD (Fig. 3B) of that point. On the other hand, the CCB and FE models assume the femur is a plate with constant thickness, and the difference in BMD is accounted for by assigning different moduli of elasticity to different points. The normal stress is therefore dependent on the material property and thus BMD (Figs. 3C and 3D), which is quite different from the CB model.

The reason why the FE model performed slightly better than the CCB model may stem from the inherent differences in FE method and CCB model theory, as well as the boundary conditions. FE method is a numerical method with the capability to tackle problems with complex geometry, material property, and loading conditions. On the other hand, the CCB model is an analytical beam theory and uses many assumptions. For a nonhomogeneous structure as complex as the proximal femur in terms of geometry and material arrangement and properties, the FE method is likely to be more accurate in representing reality than the beam theory and would certainly be consistent with our observations.

In agreement with many other studies, we showed the usefulness of HSA in discriminating fracture cases from controls. Previous studies have found that HAL is significantly longer in hip fracture than in controls,^(4,6,8,11,15,19,34) but some did not.^(10,18,35,36) Our results supported the findings that longer HAL is associated with hip fracture as a whole and with FN fracture in particular. The underlying reason for this may be the increased bending moment leading to increased stress, which was shown in our study by the significant positive correlation between HAL and IFR. Some studies^(5,10,37) showed that a larger FN

width is a significant indicator of hip fracture, which is in agreement with our results, but others^(13,15,18) did not find this association. The underlying explanation for the predictive ability of FN width, or lack of it, is unknown. If fracture occurs because of stress yield, a wider FN would protect the FN because of increased section modulus. On the other hand, if fracture occurs because of cortical buckling, a wider FN would compromise the stability because it causes an increased buckling ratio. Very recently, Rivadeneira et al.⁽³⁷⁾ showed in a prospective study that hip fracture cases in both sexes had lower BMD, thinner cortices, greater bone width, lower moment of inertia, and higher buckling ratio at base-line and that hip fracture occurred at the same absolute levels of buckling ratio. They therefore suggest that extreme thinning of cortices in an expanded FN plays a key role on local susceptibility to fracture. However, cortical buckling as a mechanism for hip fracture is still the subject of some controversy and on-going research.^(38–40)

We found some differences between FN and TR fractures in relation to the controls. FN BMD was a good indicator of FN fracture but not a good one for TR fracture compared with TH and TR BMDs. After adjusting for TH BMD, bone width was a significant discriminator of FN fracture but not for TR fracture. Differences in predictors for FN and TR fractures of the proximal femur have been reported by other researchers.^(6,15,18,21)

One strength of this study is that the cases and controls were prospectively recruited. We do, however, recognize that this study has some limitations. The controls were well matched to the cases by age, height, and weight. This is advantageous in terms of controlling study variables but restricts the extent to which the results can be extrapolated. Second, the number of subjects involved was relatively small, which may cause overfitting problems. Although we performed cross-validation checks and found that the regression models were reliable, larger studies is needed to confirm the findings. Third, we took DXA hip scans on fractured patients within 2 mo after fracture, and there is a possibility that postfracture bone loss affect the results. Jorgensen et al.⁽⁴¹⁾ reported that large bone loss occurs only in the case of a major reduction in mobility for several months. Finally this study, like all case-controlled studies, suffers from the potential effects of unrecognized sources of bias, which need to be validated by further studies, preferably longitudinal cohort studies.

In conclusion, all the IFRs derived from all the SEM models of the proximal femur were, on their own, significant discriminator of hip fracture with a 1 SD increase in the indices being associated with a 40–320% increase in the odds of hip fracture. Because of their high correlation to TH BMD, they did not improve hip fracture discrimination in the presence of TH BMD. However, the FE-derived trochanteric IFR has the potential to replace TH BMD because it performed significantly better than TH BMD in predicting all hip fractures and FN fractures.

Acknowledgments

This study was supported by the Medical Research Council (Grant 80899). The authors thank Micheal Campbell, Pete Laud, and Kevin Walter at the University of Sheffield for statistical advice and Dr Guirong Jiang, also at the University of Sheffield, for classifying hip fracture types.

APPENDIX

Curved beam model

This model was originally developed by Mourtada et al.⁽²⁴⁾ The model treats the proximal femur as two straight beams (FN and shaft) connected by a curved beam (trochanteric region). As shown in Fig. 5, the curved beam starts proximally at point e where the FN axis meets the intertrochanteric line and ends distally at point f where it has the same distance as

point e to the intersection of the neck-shaft axes. The axis of the curved beam is a circular arch and connected tangent to the FN and shaft axes at each end. The thickness of a pixel t_i (in anteroposterior direction, in cm) is determined by

$$t_i = \rho_i / \rho_b$$

where ρ_i is the pixel BMD (g/cm^2), ρ_b the BMD of the fully mineralized bone tissue that was assigned to $1.05 \text{ g}/\text{cm}^3$.⁽³²⁾ A cross-section defined by a BMD profile is composed of N pixels, and its area A , the coordinates of the centroid (x_c, y_c) and moment of inertia I can be calculated as

$$A = \Delta \sum_{i=1}^{i=N} t_i \quad (\text{in } \text{cm}^2)$$

$$y_c = \sum t_i y_i / A \quad (\text{in } \text{cm})$$

$$I_x = \Delta \sum_{i=1}^{i=N} t_i (y_i - y_c)^2 \quad (\text{in } \text{cm}^4)$$

where Δ is the spacing between adjacent pixels on the BMD profile (Fig. 5). In the regions of FN and shaft, which can be regarded as straight beam, the normal stress on a pixel of a cross-section, caused by axial force F (in N) and bending moment M (in N·cm) acting on the cross-section, is

$$\sigma_i = \frac{F}{A} + \frac{M(y_i - y_c)}{I} \quad (\text{in } 10^{-2} \text{ MPa})$$

It is obvious that the stress is proportional to the distance of the pixel to the mass centroid (neutral axis) of the cross-section. In the trochanteric region, the stress can be calculated as follows⁽⁴²⁾

$$\sigma_i = \frac{F}{A} + \frac{M(A - r_i B_i)}{A r_i (R B_i - A)} \quad (\text{in } 10^{-2} \text{ MPa})$$

where N and M are the normal force and moment at the cross-section, respectively, A is the cross-sectional area of the curved beam, R the radius of the centroidal surface, r the radius of the pixel under consideration, and B_i can be determined as

$$B_i = \sum_{j=1}^{j=i} t_j \ln \frac{r_{j+1} - r_j}{r_i}$$

A factor of safety f_i was determined for each pixel as the stress s_i divided by the yield stress s_y

$$f_i = \frac{\sigma_i}{\sigma_y}$$

The volumetric density of a fully mineralized bone was assigned as 1.89 g/cm^3 ,⁽²²⁾ and the corresponding yield stress was determined according to Keyak's equation.⁽³¹⁾

Curved composite beam model

This model assumes that the proximal femur is a curved beam with a uniform thickness in the antero-posterior direction (i.e., with rectangular cross-sections), and the differences in BMD is accounted for by different material properties at each pixel. For such composite beam, a technique has been developed to solve for the normal stress caused by bending.⁽⁴³⁾ The technique transforms the composite beam into a uniform reference beam with different cross-sectional width. An effective width of the i th sub-beam cross-section of the reference beam is defined as

$$t_i = \frac{E_i}{E_{ref}} t \quad (\text{in cm})$$

where t is the actual physical width of the beam, and E_i and E_{ref} the modulus of elasticity of the i th sub- and reference beams, respectively. From this effective sub-beam width, an effective moment of inertia I_e with respect to the horizontal (antero-posterior) axis can be calculated, and the normal stress inside the beam is determined by the beam flexure formula:

$$\sigma_i^{ref} = \frac{F}{A_e} + \frac{M(y_i - y_c)}{I_e}$$

To determine the thickness, the cross-section width of the middle FN W_n is determined from the segmented DXA image. Assuming the cross-section is circular, the cross-sectional area and moment of inertia are $\pi W^2/4$ and $\pi W^4/64$, respectively. Assuming the cross-section is a plate with a thickness t , the cross-section area and moment of inertia are Wt and $W^3t/12$. To make the area and moment of inertia the same between the circular and rectangular cross-sections, the thickness has to be $\pi W/4$ and $3\pi W/16$, respectively. An average value of $3.5\pi W/16$ was used in this study to determine the thickness.

Once the thickness is known, the volumetric BMD can be determined at each pixel as

$$\rho_i = \frac{\rho_{DXA}}{t} \frac{1.89}{1.05} \quad (\text{in g/cm}^3)$$

where the last term is the ratio of tissue density to DXA BMD.⁽²²⁾ We derived the elastic modulus E_i according to Keyak et al.⁽³¹⁾ We assigned the reference beam to have a BMD of 1.05 g/cm^2 and a volumetric BMD of 1.89 g/cm^3 and calculated the effective width t_i . The normal stress for the reference beam σ_i^{ref} was calculated in the same way as the curved beam model and converted to the normal stress of the composite model σ_i :

$$\sigma_i = \frac{E_i}{E_{ref}} \sigma_i^{ref}$$

A factor of safety f_i was determined for each pixel as the stress s_i divided by the yield stress s_y

$$f_i = \frac{\sigma_i}{\sigma_y}$$

Finite element model

This model was originally described by Testi et al.⁽²⁶⁾ In our implementation, we generated FE models for plane stress analysis with thickness, and the models were solved using ANSYS (ANSYS, Canonsburg, PA, USA). We used rectangular plane stress element PLANE42. We determined the thickness and elastic modulus the same way as the curved composite model. The Poisson's ratio was assigned to 0.35. The fall impact force was applied to the greater trochanter horizontally. The medial FN surface and the part of the medial distal femoral shaft were constrained in the horizontal direction. The nodes at the most distal femoral shaft were prevented from displacement in the vertical direction. We carried out convergence study and found that the original and half the original pixel size of the BMD map (0.54×0.6 and 1.03×1.3 mm, respectively) resulted in the same stress. We used the half original size in this study.

Idealized femur model

An idealized femur model was generated as shown in Fig. 6. It has a uniform BMD of 1.05 g/cm^2 and an FN shaft angle of 40° . The impact force applied was 1000 N. Figure 5 also shows the theoretical and calculated stresses from the CB and CCB models. The maximum relative error was 4.7%.

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FIG. 1.

Regions of interest and some of the hip structure analysis parameters used in this study. The white broken lines define the narrow ROI: narrow neck (NN), middle trochanter (NT), and narrow shaft (NS). The black lines define the femoral neck (FN), trochanter (TR), shaft (SH), and total hip (TH = FN + TR) regions of interest. The hip axis length (HAL) is the line length of a-b and the neck-shaft angle is e-c-f.

FIG. 2. The loading conditions of the two beam models (A) and the finite element model (B) that simulate a lateral fall on the greater trochanter.

FIG. 3. Typical distributions of BMD (A) and cross-sectional normal stress derived from the curved beam model (B), curved composite beam model (C), and finite element model (D).

FIG. 4.
The receiver operating characteristics (ROC) curves of the different regression models.

FIG. 5. Definitions of some variables of the two beam models (A) and the pixel thickness t_i proportional to the cross-sectional BMD profile (B). The broken white line in A shows the beam axis, the continuous line the centroid of cross-sections, R the radius of the curved beam, and r the radius of the i th pixel on the cross-section or profile.

FIG. 6. The idealized femur model and the theoretical and calculated normal stress at the lateral surfaces from the CB and CCB models.

Table 1

OR (95% CI) Associated With 1 SD Change in Parameter Values

	<i>All fracture</i>	<i>FN fracture</i>	<i>TR fracture</i>
Hip BMDs			
FN BMD	3.0 (1.8–4.9)	3.9 (1.8–8.1)	1.9 (1.0–3.7)
TR BMD	4.4 (2.5–7.7)	3.7 (1.9–7.5)	5.6 (2.0–15.7)
TH BMD	4.6 (2.6–8.3)	4.1 (2.0–8.5)	5.5 (2.0–15.2)
Hip structure analysis			
HAL	2.1 (1.4–3.1)	1.9 (1.2–3.2)	1.9 (0.9–3.8)
NN A_b	2.4 (1.5–4.0)	3.0 (1.5–6.1)	1.5 (0.8–3.0)
NN W_b	2.9 (1.8–4.9)	2.9 (1.5–5.4)	2.9 (1.2–7.0)
NN Z	1.5 (1.0–2.3)	1.9 (1.0–3.4)	1.1 (0.6–1.9)
NN t_c	4.1 (2.3–7.3)	5.2 (2.2–12.2)	2.6 (1.1–5.8)
NN r_b	3.3 (2.0–5.5)	3.4 (1.8–6.5)	2.5 (1.2–5.1)
NT A_b	5.5 (2.9–10.4)	4.4 (2.1–9.3)	7.8 (2.3–27.0)
NT W_b	3.4 (2.0–5.9)	3.3 (1.6–6.8)	3.8 (1.5–9.5)
NT Z	2.9 (1.7–4.8)	1.8 (1.0–3.3)	8.7 (2.2–33.9)
NT t_b	5.1 (2.7–9.4)	5.1 (2.3–11.5)	5.5 (1.8–16.6)
NT r_b	4.5 (2.5–8.2)	4.0 (2.0–8.0)	4.9 (1.7–14.5)
Curved beam model			
FN IFR	1.4 (1.1–2.0)	1.8 (1.1–2.8)	1.2 (0.7–2.1)
TR IFR	1.7 (1.2–2.3)	1.8 (1.2–2.8)	1.4 (0.8–2.4)
TH IFR	1.6 (1.2–2.3)	2.0 (1.2–3.1)	1.2 (0.7–2.0)
Curved composite beam model			
FN IFR	1.9 (1.3–2.8)	2.3 (1.4–3.9)	1.1 (0.7–2.4)
TR IFR	2.6 (1.7–4.0)	2.3 (1.4–3.7)	2.4 (1.3–4.6)
TH IFR	2.6 (1.6–4.0)	2.6 (1.5–4.4)	1.9 (1.0–3.3)
FE model			
FN IFR	1.8 (1.2–2.5)	2.0 (1.3–3.1)	1.1 (0.7–1.8)
TR IFR	4.2 (2.3–7.6)	2.7 (1.6–4.8)	3.0 (1.5–6.0)
TH IFR	3.2 (1.9–5.4)	2.6 (1.5–4.5)	2.0 (1.1–3.5)

NN, narrow neck; NT, narrow trochanter; A_b , cross-section bone area; W_b , bone width; Z , section modulus; t_c , cortical thickness; r_b , buckling ratio.

Table 2
Correlation Coefficients of Key Parameters of Densitometry, Hip Structure Analysis, and Structural Engineering Models

	TH BMD	HAL	NN A _b	NN W _b	NN t _c	NN r _b	NT A _b	NT W _b	NT t _c	NT r _b	CB TR IFR	CCB TR IFR	FE TR IFR
TH BMD	1	-0.01	0.78*	-0.24*	0.83*	-0.76*	0.94*	-0.10	0.92*	-0.84*	-0.55*	-0.74*	-0.78*
HAL	-0.08	1	0.17	0.34*	0.05	0.07	0.11	0.31	0.01	0.06	0.13	0.13	0.13
NN A _b	0.79*	0.13	1	0.08	0.93*	-0.74*	0.82*	0.13	0.81*	-0.62*	-0.45*	-0.55*	-0.55*
NN W _b	-0.22*	0.38*	0.10	1	-0.28*	0.55	-0.09	0.68*	-0.24*	0.42*	-0.08	0.09	0.23†
NN t _c	0.84*	-0.01	0.93	-0.27*	1	-0.90*	0.82*	-0.14	0.86*	-0.73*	-0.40*	-0.56*	-0.60*
NN r _b	-0.78*	0.16*	-0.74*	0.52*	-0.89*	1	-0.71*	0.27*	-0.76*	0.77*	0.36*	0.55*	0.65*
NT A _b	0.94*	0.07	0.81*	-0.07	0.81*	-0.71*	1	0.09	0.92*	-0.80*	-0.59*	-0.72*	-0.75*
NT W _b	-0.13	0.42*	0.12	0.64*	-0.13	0.27*	0.06	1	-0.10	0.27*	0.18	-0.03	0.10
NT t _c	0.93*	-0.06	0.79	-0.19*	0.83*	-0.74*	0.94*	-0.12	1	-0.99*	-0.52*	-0.67*	-0.71*
NT r _b	-0.86*	0.18*	-0.62*	0.33*	-0.71*	0.78*	-0.81*	0.30*	-0.86*	1	0.51*	0.68*	0.79*
CB TR IFR	-0.57*	0.27*	-0.58*	-0.07	-0.44*	0.52*	-0.56*	-0.06	-0.53*	0.56*	1	—	—
CCB TR IFR	-0.75*	0.28*	-0.66*	0.12	-0.59*	0.72*	-0.71*	0.09	-0.70*	0.77*	0.77*	1	—
FE TR IFR	-0.72*	0.27*	-0.61*	0.19*	-0.56*	0.77*	-0.66*	0.15†	-0.64*	0.81*	0.81*	0.81*	1

Coefficients in the lower and upper triangles are for all fracture + control and FN fracture + control, respectively.

Correlation is significant at the * 0.01 † 0.05 levels, respectively.

NN, narrow neck; NT, narrow trochanter; A_b, cross-section bone area; W_b, bone width; Z, section modulus; t_c, cortical thickness; r_b, buckling ratio.

Table 3

Sensitivity at 80% Specificity (Specificity at 80% Sensitivity) of Various Regression Models

<i>Regression model</i>	<i>Original regression</i>	<i>Bootstrap cross-validation</i>
All hip fracture and control		
TH BMD	50 (58)	50 (59)
TH BMD + NN width	67 (72)	67 (72)
FE TR IFR	61 (70)	61 (69)
Femoral neck fracture and control		
TH BMD	45 (53)	42 (54)
TH BMD + NN width	60 (68)	54 (67)
FE TR IFR	58 (70)	53 (66)
Trochanter fracture and control		
TH BMD	56 (67)	56 (68)
FE TR IFR	61 (67)	66 (67)