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Method for Cortical Bone Structural Analysis from Magnetic Resonance Images

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Abstract

Rationale—Quantitative evaluation of cortical bone architecture as a means to assess bone strength is typically accomplished on the basis of images obtained by dual-energy X-ray absorptiometry (DXA) or computed tomography (CT). Magnetic resonance imaging (MRI) has potential advantages for this task in that it allows imaging in arbitrary scan planes at high spatial resolution. However, several hurdles have to be overcome in order to make this approach practical, including resolution of issues related to nonlinear receive coil sensitivity, variations in marrow composition and the presence periosteal isointense tissues which all complicate segmentation.

Objectives—To develop MR acquisition and analysis methods optimized for detection of cortical boundaries in complex geometries such as the femoral neck.

Materials and Methods—Cortical boundary detection is achieved by radially tracing intensity profiles that intersect the bone's periosteal and endosteal boundary. The profiles are subsequently normalized to the intensity of the marrow signal, processed with morphologic image operators and binarized. The resulting boundaries are then mapped back onto the spatial image and erroneous boundary points removed. From the detected cortical boundaries cortical cross-sectional area and thickness are computed. The method was evaluated on cortical bone specimens and human volunteers on the basis of high-resolution images acquired at 1.5 Tesla field strength. To assess whether the method is sensitive to detect the expected dependencies of cortical parameters in weight-bearing bone on overall habitus, ten women ages 46–73 years (mean 56 years) underwent the cortical imaging protocol in the proximal femur and the results were compared with DXA BMD parameters of the hip and spine.

Results—Reproducibility was on the order of 2%. Double oblique images of the femoral neck in the ten women studied showed cortical cross-sectional area to correlate strongly with height ($r = 0.88$, $p = 0.0008$) while cortical diameter versus age approached significance ($r = 0.61$, $p = 0.06$). Measurements in specimens of some cortical parameters indicated a resolution dependence. It is to be noted, however, that parameter rank remained constant across all specimens studied.

Conclusion—The data suggest the new method to be robust and applicable on standard clinical MR scanners at arbitrary anatomic locations to yield clinically meaningful quantitative results.

Indexing

Cortical bone; computer-aided measurement; magnetic resonance imaging; osteoporosis

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Introduction

In age-related and postmenopausal osteoporosis, bone loss leads to both trabecular bone atrophy and cortical thinning (1–3). This effect is partially compensated by an increase in cortical diameter (4), thus increasing torsional stiffness while reducing buckling resistance (5). Recent data suggest that weakened cortical bone may primarily be responsible for hip intracapsular hip fracture (6), thus emphasizing the importance for quantitatively assessing and monitoring cortical structure. Support for the role of cortical bone architecture as a risk factor for hip fracture is found in several studies of postmenopausal osteoporosis, where strong associations have been reported between proximal femur geometry and fracture incidence (7–9). Additionally, bone mineral density (BMD) supplemented by femoral geometry has been shown to be more predictive of breaking strength than BMD alone (10).

The noninvasive assessment of cortical bone structure in osteoporosis has typically been performed on the basis of x-ray projection radiographs (11) and DXA BMD images (9,12). However, the projection nature of these images ignores the 3D aspect of cortical architecture. The femoral neck axis is angulated relative to the coronal plane, therefore making it difficult to accurately evaluate cross-sectional geometry from such images. Computed tomography (CT) (13,14) and peripheral quantitative CT (pQCT) (15,16) have been used to measure 3D cortical geometry, but these techniques are confined to the more distal locations of the peripheral skeleton, such as the wrist and ankle. Nevertheless, these studies support the importance of 3D cortical structure analysis. MR is uniquely suited for direct acquisition of images at arbitrary orientation to optimally capture this 3D structure. By contrast, CT would require reformation of axial data, typically resulting in a loss of resolution.

Recently, a number of articles have appeared in which quantitative MRI approaches were used for quantifying cortical bone structure and geometry (17–25) and for musculoskeletal applications in general (26). For example, Murdoch *et al* (20) showed an association between humeral length and height in a group of 20 volunteers ($r=0.79$, $p<0.05$). Duncan *et al* (21) showed mid-femoral cortical structure and BMC differences between exercise categories using MR. Wehrli *et al* (25) were able to use cortical bone measurements to assess the consequences of renal osteodystrophy, showing reduced cortical cross-sectional area and thickness in patients relative to matched controls ($P = 0.008$ and $P = 0.01$ respectively). These studies highlight the potential of MRI for in vivo cortical bone structure analysis.

In this work, a MR image-based processing algorithm was developed for cortical bone structure analysis. It is based on transecting the cortical bone along test lines (profiles) to locate the cortical boundaries along each profile. The goal was to develop a robust method for cortical bone analysis from MR images applicable to various clinically relevant anatomic sites. The method has been evaluated with experiments estimating resolution dependence and precision from repeat volunteer scans. The method's potential is further highlighted with the results of a pilot study investigating whether the method is sensitive enough to measure the expected dependence of femoral architecture on body habitus. Preliminary accounts of this work have previously been given in abstract form (27,28).

Materials and Methods

Image Processing

MR-based cortical bone analysis is complicated by bone's signal characteristics and anatomy. Bone is essentially a solid, which in MR images appears with background signal intensity in spite of a 10–15% water content which has $T2<1\text{ms}$ (29) and therefore is ordinarily not detectable, while the medullary space contains various proportions of hematopoietic and fatty marrow. The periosteal region consists of various soft tissues including muscle and connective

tissues (e.g. tendon) which, similar to bone, appear with background intensity by virtue of their extremely short T2. This make-up produces pulse-sequence-dependent contrast along the endosteal and periosteal boundaries. The detection of the endosteal boundary is further complicated by partial volume effects from the often gradual transition from trabecular to cortical bone. On the periosteal boundaries, the presence of low-intensity connective tissue such as tendons and ligaments adjacent to cortical bone, unless appropriately dealt with, could mistakenly be assigned to cortical bone. Finally, the chemical shift artifact between fat and water (30) causes a spatial shift in the images along the frequency-encoding direction that must be compensated for. The approach here addresses these problems to yield reliable boundary detection.

Brief Description of Algorithm

The first step consists of generating a region defining the marrow, which roughly follows the cortical boundary. The boundary of this region is then used to trace outward-directed profiles of sufficient length to transect the cortical bone. By following the cortical shell around the circumference of the bone at small angular increments, these profiles transect the boundaries sequentially, producing a profile map. The profiles are then normalized to the intensity of the marrow signal, which is considered to be the highest intensity among all tissues in the image. The profile maps are subsequently processed with one- and two-dimensional morphologic image operators and binarized to determine the cortical boundaries. Finally, the boundaries are mapped back onto the spatial image and erroneous boundary points removed. From the detected cortical boundaries cortical cross-sectional area and thickness are computed.

Marrow Region Detection

To accurately locate profiles that transect the cortical bone, a rough estimate of the marrow region is generated. This marrow region then serves as the basis for selecting the starting locations for profile extension. At locations such as the tibial midshaft, the cortical bone is thick enough so that a simple 3D region-growing technique can accurately isolate the marrow region. Toward this goal a seed point is manually selected and the neighboring intensities used to determine the marrow intensity mean and standard deviation (SD). To reduce the noise in the image, an 11-voxel square median filter is applied, a size small enough not to span the cortical shell. The region growing proceeds from an arbitrarily selected central seed point to include all connected voxels at intensity range of within ± 2 SD of the marrow average. Typically, a 95% confidence interval is used (± 2.6 SD), but for this application the marrow region should be slightly smaller than the endosteal surface, so ± 2 SD was found to give better results.

At locations rich in trabecular bone (e.g. the distal tibia) the cortical shell is thinner and the use of surface coils leads to intensity shading, thus requiring a modified processing algorithm. In these cases, image intensities are first normalized to eliminate low-frequency intensity gradients (Figure 1a). Assuming that fatty marrow voxels have the highest intensity and that these tissues are evenly distributed throughout the image, a 30-pixel-radius circular processing window is passed over the complete image and the 95th-percentile ranked intensity is used as the estimate for marrow intensity. This value filters out high-intensity outliers and produces an image of normalized intensities used to adjust image intensity inhomogeneity (Figure 1b) by division of the original intensity by the normalization value. To isolate the marrow region automatically, a series of image processing operations is performed on the normalized image:

1. a 3D morphologic gray-scale opening operation (Figure 2a) to disconnect the marrow space from the surrounding soft tissues;

2. binarization using a fixed threshold of 128 corresponding to 50% of the marrow intensity;
3. placement of an operator-selected seed point in the marrow space as the starting location for 3D region-growing (Figure 2b);
4. a 2D morphologic binary closing operation;
5. background region-growing to isolate the marrow region, and image inversion;
6. edge smoothing using another 2D opening operation to yield the marrow boundary (Figure 2c).

At sites such as in the femoral neck where the cortex is thin, the above approach may fail. However, it typically suffices to establish an approximate endosteal boundary since this process merely serves to determine the starting locations for tracing profiles. The only requirement for this is to be close enough to the cortical bone so that the profile can intersect both endosteal and periosteal boundaries.

Profile Location and Boundary Detection

Following rough outlining of the marrow cavity, profiles of length sufficient to span the cortical bone are specified (Figure 3). The intensities along each profile are subsequently arranged consecutively in an array (profile position versus profile number) for detection of endosteal and periosteal boundaries. Toward this goal, the marrow region is first morphologically eroded by about one quarter the profile length to separate the starting locations of the profiles from the endosteal boundary. The region border is detected by subtracting the same region morphologically eroded by one voxel and the orientation of the boundary is determined at each location by least-squares fitting the five neighboring boundary locations to a straight line. Profiles are then extended from the boundary location in an outward direction perpendicular to the fitted line so as to traverse the cortex (Figure 3, solid lines). The MR image intensities along each profile are sampled from the normalized image, and the profiles arranged consecutively creating a 2D array (Figure 3 dashed box subset shown in Figure 4a original, **b** normalized), termed a profile map. It should be emphasized that the profiles thus generated cross the cortical bone in a very dense manner so as to sample the cortical boundaries in as many locations as possible.

To determine the cortical boundary locations, the profile map is processed by a 3 x 3 median filter to remove noise along and across profiles (Figure 4b). The map is binarized at an empirically determined threshold of 30% of the maximum intensity (Figure 4c), which corresponds to a value sufficiently low to detect the periosteal boundary at the bone muscle interface. A morphologic closing operation along each profile is used to remove noise and artifacts from within the cortical bone, such as from blood vessels. The endosteal boundary of each profile is then determined as the first nonzero pixel and the periosteal boundary as the first zero pixel after that. Finally, the cortical boundary points determined along each profile are mapped back to the original MR image and the endosteal and periosteal boundaries are thus located (Figure 4d and e).

Removal of Erroneous Boundary Points

Because some tissues surrounding the bone do not produce a detectable MR signal, the boundary point detection sporadically misses the endosteal or periosteal boundary, resulting in erroneous boundary points. Because the boundaries are highly oversampled, only a fraction of the points detected are needed for segmentation and thus erroneous points can be removed without loss of accuracy. The method for removing erroneous cortical boundary points consists of a three-step algorithm. In the first step erroneous boundary points at the extreme ends of the

profile (as these indicate truncation errors) are removed. The second algorithm first computes the distance from each boundary point to its neighboring endosteal and periosteal points. If an endosteal point is closer to the outer cortical boundary, it is discarded, and vice versa in the periosteal case.

The third algorithm is based on the notion that cortical boundaries are smooth with only small variations in curvature. The algorithm removes erroneous boundary points by searching for sharp spatial discontinuities, which is achieved by analyzing the change in orientation along the boundary. For example, when moving clockwise along the cortical boundary, and encountering a 90° left turn, a jump, a 90° right turn, followed by a number of boundary points, and a subsequent reverse 90°/jump/90° sequence, the cortical boundary detection is likely in error and the group of distant boundary points are removed. To allow for variability in detecting erroneous points, allowable deviation from 90° turns is set to 10°, (corresponding to 80° – 100°), the length of the jump to five pixels minimum, and the maximum number of points in the distant group to 10. This algorithm removed approximately 95% of all erroneous points and in most cases produced a smooth boundary. To remove undetected erroneous boundary points a mechanism was put in place to allow manual deletion of points. The high density of profiles along the cortical boundary oversamples the boundary ensuring that after the removal of erroneous points there are a sufficient number of ‘good’ points to detect the complete cortical boundary and calculate accurate cortical structural parameters.

Chemical Shift Artifact Correction

The chemical shift effect in MRI causes a spatial displacement of the fat and water signal owing to their different resonance frequencies (31). Since the marrow is partially (femur, spine) to mostly fatty (distal radius, tibia) and the tissues abutting the periosteal boundary are mainly water (muscle) this effect can be substantial. Even in the case of partially fatty marrow, the signal intensity from the fat is much higher and the boundary detected essentially represents the fat. This effect needs to be corrected by shifting the detected endosteal boundary relative to the periosteal one by the appropriate number of pixels in the readout gradient direction (chemical shift (Hz) divided by the bandwidth per pixel). After this correction, chemical shift-compensated cortical segmentation images were used for all further analyses.

Cortical Parameters

The corrected cortical boundaries are traced as polygons, defining the cortical bone and medullar segmented image. The cortical area (area between endosteum and periosteum) and bone area (encompassed by the periosteum) are computed as the sum of pixels multiplied by the pixel area. Cortical mean thickness (CMT) is calculated by modeling the endosteal and periosteal boundaries in each image as concentric circles whose radii are derived from the respective areas. Horizontal neck width and vertical neck height were measured from femoral neck images as the distance between the farthest endosteal boundary points on the x- and y-axes, respectively. Femur axis length and hip axis length were computed using standard definitions (7) as shown in Figure 5 but the measurements were taken from the endosteal boundary determined by the boundary detection method described above.

Clinical Imaging

The method was evaluated at two different skeletal locations: the tibia and femur. Reproducibility was assessed using test-retest experiments, and possible resolution dependence was evaluated in the tibia using human tibia samples. Measurements at the tibial midshaft were motivated by the effect on cortical structure of some metabolic bone diseases, such as Gaucher’s disease (32) and hyperparathyroidism (33,34), disorders that are known to primarily affect cortical bone (35).

Finally, the cortical analysis methods were applied to a small pilot study of cortical structure at the proximal femur and femoral neck in a group of healthy female volunteers for evaluating the changes of these parameters relative to anthropometric measures. The study subjects also underwent a protocol for DXA imaging of the hip and spine using a Hologic QDR4500a device (Hologic, Inc., Bedford, MA). Measurements were obtained of the anterior-posterior lumbar spine, and the right and left proximal femur in array mode using standard positioning techniques.

All cortical imaging protocols were executed on a General Electric Signa™ 1.5 Tesla clinical scanner (Milwaukee, WI, USA) with Epic 5.6 console, Echosped gradients, and acquisition protocols specially developed for reproducible image localization.

For cortical analysis of the distal tibia, the body coil was used to obtain a sagittal image of the complete bone (for bone length measurement). Subsequently, the General Electric 6.5" diameter extremity coil (model 46-265442G1) was used for high-resolution sagittal localizers and axial cross-sectional images equally spaced along the distal half of the tibial diaphysis. For imaging tibial cortical bone, a 2D fast spin-echo (FSE) scan produced about 30 axial images at an in-plane pixel size of $0.47 \times 0.47 \times 2 \text{ mm}^3$. Table 1 shows the acquisition parameters for this protocol.

Cortical analysis at the femoral neck, the most common site of osteoporotic hip fractures, is complicated by its deep location within the pelvis and the double-oblique angle between the neck and the body's orthogonal axes. Consequently, an imaging protocol was chosen to produce images for the complete measurement of geometric parameters at the proximal femur, previously reported in preliminary form (28). Because the proximal femur has a complex 3D shape, these measurements were done on oblique slices perpendicular to the bone's anatomic axes. Images of the proximal femur were acquired using the General Electric four-element Cardiac Array coil placed around the hip. After an initial gradient-echo (SPGR) axial localizer to locate the femoral axis, oblique 2D FSE slices were acquired across the proximal femur parallel to the femoral axis in the antero-posterior direction for geometric measurements. The central slice was determined as the one with the largest neck diameter and subsequently double-oblique femoral neck GE-SPGR images were collected along the length of the neck to measure cortical bone. MR imaging parameters were optimized for visualization of the cortical boundaries in each imaging plane and are summarized in Table 1. The cortical boundaries of the MR images were located as described previously (Figure 6).

Assessment of Reproducibility

To assess the method's reproducibility at the tibia and femur, two healthy volunteers were scanned repeatedly. At the distal tibia diaphysis, reproducibility was tested with images from two subjects (male, 37 year old, Caucasian; and female, 35 year old, black), each scanned three times with repositioning between scans. For measurements of femoral parameters, two subjects (female, 35 year old, Caucasian; male, 29 year old, Asian) were scanned three times each in independent imaging sessions. The intraclass correlation coefficient (ICC) was estimated from the SD within and across subjects and the coefficient of variation (CV) from the within subject SD and the across subject average. ICC was calculated from the F statistic for across-subject to the average within-subject variance ($ICC = (F-1)/(F+k-1)$ where k is the number of tests) (36).

Localization and angulation of the femoral neck cross-sectional images was achieved by visual inspection of the oblique coronal images of the proximal femur. Ideally, the image plane would be exactly parallel to the direction of the femoral neck at the location of its minimal diameter. However, in practice there may be a small error in selecting this plane (which is performed by visual inspection). To correct for such small angle variations in the tilt of the cross-sectional

images (from which cortical structural parameters are derived) relative to the neck axis, the vertical neck height, defined above, was measured in both the axial (Va) and coronal (Vc) images, and subsequently the cross-sectional parameters corrected by the factor V_c/V_a .

Resolution Dependence

To assess the resolution dependence, six tibia cortical specimens harvested at autopsy from three subjects (46, 84, and 86 years old) were scanned at multiple resolutions, from $0.137 \times 0.137 \times 2 \text{ mm}^3$ to *in vivo* voxel size of $0.470 \times 0.470 \times 2 \text{ mm}^3$. Multiple repeat scans were averaged at in-plane resolutions of 0.137 mm (three times), 0.313 mm (three times), and 0.391 mm (two times). Specimens of human cortical bone were sectioned from the tibial midshaft at locations 5 and 10 cm from the distal endplate. The specimens were immersed in saline and scanned with a FSE sequence (TE=19 msec, TR=2 sec, ETL = 8) and a custom-built wrist coil for adequate signal-to-noise at the higher resolutions. The cortical parameters compared were bone area (BA), cortical area (CA), normalized cortical area (CA/BA) and cortical thickness. The highest resolution was considered as reference (“gold standard”) and was used to compute errors at the other resolutions.

Femoral Neck Pilot Study

A pilot study was conducted to examine the method’s sensitivity to detect the known dependencies of cortical bone architecture at the femoral neck on anthropometric measures. An additional motivation was to determine whether the detected changes were significant enough to require normalization when used in future studies as a metric to evaluate the implications of cortical bone on hip fracture incidence. Toward these goals ten women, ages 46–73 years (mean 56 years), were randomly selected from a prior study (37). The subjects, who had DXA BMD T-scores of the femoral neck of -1.19 ± 1.39 (mean $\pm$ SD) underwent the cortical hip analysis protocol described above.

Results

Reproducibility

Reproducibility in the tibia and femur was assessed from measurements of two volunteers, scanned three times each, as described previously. Figure 7 shows the test–retest plots in the tibia, with regression analysis for CA along all slices of one subject. When combining the two subjects’ data, ICC and CV (%) values for CA were 0.78 and 1.78 %, for cortical mean diameter 0.994 and 0.69 %, and for cortical mean thickness 0.98 and 2.03 %, respectively. In the proximal femur, average CV across all parameters was 2.4% and 1.4% for femoral neck parameters and hip geometry respectively. Femoral parameter ICC estimates ranged from 0.79 to 0.98, with an average of 0.86.

Resolution Dependence

The resolution dependence experiments indicate a nearly linear increase of the apparent structural parameters with image pixel size (average r^2 values ranging from 0.89 to 0.92; Figure 8). It is evident from Figure 8 that the slopes are very similar and that the rank of the parameters is preserved. The error at *in vivo* resolution relative to the “gold-standard” resolution is 6% for bone area, and 29% for cortical thickness, the latter being particularly difficult to measure at *in vivo* resolution. Figure 8 shows a representative graph of cortical area versus resolution, illustrating the linear dependence and the rank preservation. The anatomic accuracy of boundary detection is illustrated with the data from the femoral neck and tibia example shown previously (Figure 6).

Femoral Neck Pilot Study

The results from the femoral neck pilot study show that for correlation with age, no parameter was significant; however several MR-derived geometric parameters approached significance, with neck diameter leading above the rest ($r = +0.61$, $p = 0.06$). DXA BMD correlated significantly with the subjects' weight yet MR parameters correlated significantly with the subjects' height and weight (Table 2). Interestingly, correlations involving height were generally stronger than those for weight. Cortical area and average thickness correlated only with height. The parameter with strongest correlation with height was cortical area ($r = 0.88$ and $p = 0.0008$, Figure 9).

The average correlation between hip geometry and MRI femoral neck parameters (bone and cortical area) was 0.59 but the structural parameters were more weakly correlated with BMD parameters ($r = 0.32$). BMD parameters correlated with femur axis length but not hip axis length. MR-derived femoral neck cortical parameters, on the other hand, correlated more strongly with hip axis length than femur axis length (Table 3).

Discussion

The strength of the approach described in this work is its ability to image thin, complex 3D cortical structures of the most common sites of fracture, i.e. the femoral neck, distal radius, and tibia. Among the difficulties to be overcome are chemical shift displacement artifacts, the potential confounds of isointense adjacent structures, and erroneous boundary points. The method offers a high degree of automation requiring little operator intervention.

The validation studies show excellent *in vivo* reproducibility. The ICC and CV parameters from the reproducibility experiments indicate the method to be precise for analyzing cortical bone cross-sectional parameters. The CA parameter shows low ICC but also very low CV. Overall, the regression analysis, ICC and CV measures indicate that the method is reliable with an estimated error of about 2%.

Apparent cortical structural parameters were found to be dependent on image resolution and the present study was not designed to evaluate absolute accuracy, which would require comparison with anatomic sections. Nevertheless, if we regard the highest resolution achievable *in vivo* ($137 \times 137 \mu\text{m}^2$ pixel size) as a "gold standard", the data suggest a systematic error that increases with increasing resolution, most likely the result of partial volume effects. This error is very consistent as indicated by the fit lines of Figure 8. It is noticeable from these data that the slopes are almost parallel and thus the rank does not change as a function of resolution. We thus conclude that comparisons are valid at a given resolution, and most likely errors are similar across a range of structural values. To further test this hypothesis a more extensive study of resolutions would be needed between two groups of samples with known differences. Such a study would show if statistical power in discriminating between groups is preserved across resolutions.

Our data confirm significant correlations between femoral geometry and anthropometric parameters (notably weight and height). Similar findings were previously reported using DXA projection techniques (7), however the associations were much weaker in their study. The observed relationships are expected as it is well established that the skeleton scales with height and weight (38,39). Femoral neck width increasing trend seen with age ($r = 0.606$; $p = 0.063$) found in our study are consistent with previously reported studies (4,5) which suggesting subperiosteal expansion and cortical thinning with age.

Besides absence of ionizing radiation, MR has several advantages over quantitative computed tomography. Oblique femoral MR images eliminate the limitations and error sources of

projection imaging techniques, and double oblique images across the femoral neck prove useful for evaluating cortical neck cross-sectional parameters. Another motivation for the use of MR to analyze cortical bone is its ability to capture and analyze both trabecular and cortical bone as seen in a previous study on patients with renal osteodystrophy (25) thus enabling comprehensive assessment of both cortical and trabecular bone architecture by a single modality.

The cortical boundary analysis presented here assumes that the location of the first high-intensity pixel outside the marrow cavity represents periosteum. This assumption may not be satisfied in the case of high-intensity intracortical structures (e.g., arteries, veins) and low-intensity extracortical structures (e.g., tendons, ligaments). The former is not a significant problem since intracortical vessels are usually small and ordinarily below the image resolution limit. However, surrounding tendons and ligaments, both soft tissues with high collagen and low water content, may have very low intensity except at echo times much shorter than tissue T2. This confound, potentially leading to erroneous boundary detection, can be circumvented by using short-TE imaging sequences. Further, the choice of intensity threshold used for detecting the cortical boundaries is based on the much lower intensity of muscle tissue surrounding the bone. In future applications of the technique, this threshold could be selected separately for endosteal and periosteal boundary detection, depending on the tissue intensities actually present along each profile.

In summary, our data indicate that MRI is suited for analyzing cortical bone structure, offering an alternative to existing x-ray-based methods. Enhancement of the spatial sensitivity of a dedicated hip coil would further improve the performance of the method in the proximal femur. Finally, our data indicate substantial sensitivity in detecting the implications of body weight and height on cortical bone architecture of the hip, which underscores the need for data normalization when cortical structure is used as a possible indicator of hip fracture risk.

Conclusions

This work shows that parameters related to cortical bone architecture can be derived from MR images. The method exploits the high contrast between bone and soft tissues allowing reliable detection of periosteal and endosteal boundaries. The data presented show the method to be reproducible and robust, requiring minimal user intervention.

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Figure 1.

a) Unprocessed MR image of the distal tibia (137x137x410 microns using 3D spin echo sequence); b) image after intensity normalization. Notice the strong shading effect in images acquired using the two-element surface coil.

Figure 2.

Sequence of automated operations to determine the marrow region (distal tibial metaphysis) :
a) gray-scale opening operation separating the marrow space from surrounding soft tissues; b)
marrow space selected by 3D region-growing; c) overlay of detected boundary on original
image.

Figure 3. Principle of the intensity profile-based boundary detection algorithm illustrated on the distal tibial metaphysis. The dotted line represents the curve of points from which profiles are drawn outward. Solid lines represent evenly spaced profiles along this curve, with each profile transecting the cortical shell. The rectangular box encompasses a region whose processing steps are shown in subsequent figures.

Figure 4.

Steps of the cortical boundary detection algorithm: a) Display of adjacent profiles from the designated region in Figure 3. The bottom edge is the region within the trabecular bone and shading across the profiles is clearly seen. b) Intensity-normalized image with no shading clearly showing morphologically closed marrow region and cortical bone layer (low-intensity

band). (c) Filtered and binarized cortical image. (d) Endosteal and periosteal boundaries shown in red and green, respectively, on the intensity-normalized profiles. (e) Region of the original MR image with the endosteal and periosteal boundaries highlighted.

Figure 5. Coronal oblique FSE image of the upper femur parallel to the femoral neck axis. Measurable geometric quantities: ND = neck diameter, HD = head diameter, NA = neck angle, hip axis length and femur axis length.

Figure 6. Examples of femoral neck (a) and tibia (c) images with resulting automatically located boundary points (b and d resp.).

Figure 7.

Regression analysis of test–retest measurements of cortical area. Tests 2 and 3 are plotted against those of test 1 with linear regression analysis results indicated on the graph.

Figure 8.

Plot of cortical to bone area ratio versus in-plane resolution at several resolutions, some with repeat measures. Linear trends show a significant increase in parameter measurement with resolution. Notice that due to the parallel nature of the linear correlations the rank of the subjects does not change across resolutions.

Figure 9. Femoral neck cortical area versus subject height showing a highly significant correlation. Ten subjects were evaluated using a spin echo T1-weighted MRI sequence and the femoral neck parameters compared with patient weight, height and age.

Table 1

MR Scan Protocol Parameters

Scan	FOV (cm)	TR (ms)	TE (ms)	THK (mm)	skip (mm)	time (min)	NEX	Phase enc.	ETL
Tibia diaphysis	12	5000	16	2	5	2:42	1	256	8
Femoral localizer	20	68	2.2	4	2	0:27	1	128	
Femoral geometry	16	4000	16	2	1.5	4:16	2	256	8
Femoral neck	16	150	4.8	2	0	5:10	4	256	

Table 2

Pilot study parameter correlation with age, weight and height: correlation coefficients and p values (in parentheses).

Parameter correlated	Age	Weight	Height
DXA Left neck BMD	-0.250 (0.490)	0.63 (0.05)	0.62 (0.06)
DXA Left inter-trochanteric BMD	-0.049 (0.89)	0.80 (0.006)	0.60 (0.07)
femur axis length	0.38 (0.27)	0.83 (0.003)	0.62 (0.054)
hip axis length	0.52 (0.13)	0.46 (0.18)	0.55 (0.1)
neck width	0.61 (0.06)	0.48 (0.16)	0.45 (0.19)
head width	0.4 (0.22)	0.87 (0.001)	0.73 (0.017)
femoral neck bone area (BA)	0.52 (0.130)	0.50 (0.14)	0.60 (0.07)
femoral neck cortical area (CA)	-0.04 (0.91)	0.40 (0.25)	0.88 (0.0008)
femoral neck cortical thickness	-0.36 (0.310)	0.19 (0.6)	0.64

Table 3

Correlation coefficients for average parameter correlations between DXA BMD and MRI-derived structural parameters in the femoral neck (bone and cortical area) and hip geometry (p values in parentheses).

Parameter correlated	BMD	Cortical parameters
hip geometry	0.32 (0.38)	0.59 (0.12)
hip axis length	0.39 (0.32)	0.62 (0.06)
femur axis length	0.58 (0.10)	0.54 (0.10)