Noninvasive Femur Bone Volume Estimation Based on X-Ray Attenuation of a Single Radiographic Image and Medical Knowledge

Supaporn KIATTISIN*†, Student Member and Kosin CHAMNONGTHAI†, Member

SUMMARY Bone Mineral Density (BMD) is an indicator of osteoporosis that is an increasingly serious disease, particularly for the elderly. To calculate BMD, we need to measure the volume of the femur in a noninvasive way. In this paper, we propose a noninvasive bone volume measurement method using x-ray attenuation on radiography and medical knowledge. The absolute thickness at one reference pixel and the relative thickness at all pixels of the bone in the x-ray image are used to calculate the volume and the BMD. First, the absolute bone thickness of one particular pixel is estimated by the known geometric shape of a specific bone part as medical knowledge. The relative bone thicknesses of all pixels are then calculated by x-ray attenuation of each pixel. Finally, given the absolute bone thickness of the reference pixel, the absolute bone thickness of all pixels is mapped. To evaluate the performance of the proposed method, experiments on 300 subjects were performed. We found that the method provides good estimations of real BMD values of femur bone. Estimates shows a high linear correlation of 0.96 between the volume Bone Mineral Density (vBMD) of CT SCAN and computed vBMD (all P < 0.001). The BMD results reveal 3.23% difference in volume from the BMD of CT SCAN.

The authors are with the Department of Electronics and Telecommunication Engineering, King Mongkut's University of Technology Thonburi, Prachachuen Road, Bangkok 10140, Thailand.

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a) E-mail: supaporn.kai@utcc.ac.th
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using vBMD may more accurately reflect bone density [7]. Estimated volumetric Bone Mineral Density at the Femoral Neck (vFNBMD) is more sensitive than areal Bone Mineral Density at the Femoral Neck (aFNBMD).

Comprehensive three-dimensional (3D) information about the tissues can be obtained by Computerized Tomography (CT) imaging. The basic philosophy of CT is to use a large scanner to measure an extensive set of projections from full view angle for the 3D reconstruction. Reconstruction [8], [9] quality is extremely good, and CT scans are an indispensable part of radiology today. However, QCT imaging has some disadvantages such as high radiation dose, high equipment cost, international recognized training course. Therefore, the use of CT has been mainly limited to diagnosing serious diseases and it is typically utilized in large hospital units. Unfortunately, QCT and DXA machines are not often found in hospital equipment, while radiographic devices are widely available. Consequently, the applicability of the system to conventional radiographic images was tested. A diagnostic method for telling whether a patient is suffering from osteoporosis or not is very important. Such a method makes it possible to start a treatment that can increase the bone mineral density to a level where the risk of fractures is reduced. Several such methods exist, some of them being based on x-ray image.

2. Related Work

Imaging techniques are widely used in medical practice. They have become an important tool in many areas, such as surgery planning and simulation, intra-operative navigation, radiotherapy planning, and tracking of the progress of diseases, etc. As a result, a lot of research work has been done in computer-aided medical image analysis. One method that has been extensively researched is reconstructing 3D bone models from bplanar x-ray images. Many of these 2D-3D registration techniques have focused on detection. Manually picking landmarks can provide accurate results in certain cases, but is too simplistic an approach for most applications as it can be sensitive to noise and the shape of the contour, especially in applications like the spine [10]. An obvious solution to this problem is incorporating active shape models and active appearance models [12]. Much of the work in this area, has focused on simplified shape models that only incorporate major landmarks or other prominent features on the bone. Deformable models have many ways such as active contour [11], active shape [12], and level set method [13] have also been used for medical image segmentation [15]. These methods are contour-based instead of region-based. So, they have the potential of extracting contours of body parts that do not contain homogeneous features. An important weakness of these approaches is that they typically require good initialization of the model contour. If the model is poorly initialized, these approaches can be easily affected by noise and extraneous edges in the image, resulting in incorrect segmentation of the regions.

This research concentrates on the method of constructing three-dimensional volume from two-dimensional image. There are several techniques. Some techniques use multiple 2D images while some techniques use a single 2D image for construction. The obtained image must be taken by radiation. If the body receives too much radiation then the body has a reaction. So we are interested to measure BMD from radiography.

3. Basic Principle

The 3D reconstruction from a single two-dimensional (2D) image can be done using the principle of x-ray attenuation because each object has a particular capability of ray absorption. Different objects with different depth can absorb different amounts of x-rays. Normally, the x-ray through the objects with different size has different intensity. The object size is proportional to the image intensity.

Figure 1, the x-ray with intensity is through objects with different sizes X and 2X. The object of size 2X is 2 times less capable than that of size X to let the ray go through it. This results in the intensity of the object of size 2X to have lower intensity than that on the object of size X. The x-ray intensity transmitted through a dense material is given by:

$$I = I_0e^{-\mu t}$$

Where $I$ is the transmitted x-ray intensity, $I_0$ is the incident x-ray intensity, $\mu$ is the linear attenuation coefficient (in cm$^{-1}$) and $t$ is the thickness of the material (in cm). This equation shows that the x-ray intensity depends on the density of the material and the thickness of the material. From above equations, the relative depth of each pixel can be computed rather than the absolute depth. However, if we know the absolute depth of at least one pixel, we can also find the absolute depth of all other pixels in the image.

Figure 2 shows an example of the spherical-shaped object which the x-ray passes through. The brightness of a film receptor is related to the depth of the object. If the diameter of the object can be computed, the depth of the center point can be also calculated. Then, using the center point as a reference, we can find the relative depth of other pixels in the image.

Using above knowledge illustrated in Fig. 3, we can obtain the volume of the femur. Some medical research [16]
4. Methods

The goal of this research is to estimate vBMD (g/cm³) value from the volume of human femur. Figure 4 illustrates the proposed method. The first step is the femur extraction. First of all, we must enhance the x-ray image for detecting boundary of the femur bone. We want to extract only a femur without the hip. The second step is important for finding a reference point. In this case, we consider a femur head which looks like a sphere. If we know the center point of circle, we can obtain a diameter and intensity at midpoint. The midpoint of diameter is first reference point as well. The value of that point can represent with each point of region of interest (ROI). This step is semi-automatic step because it depends on midpoint which we obtain. In this method, the midpoint is the first reference point or the reference point that we must decide again. The new point uses data of the first midpoint for finding intensity and a depth at the new point. Thus the new point is the reference point. The third step is finding a volumetric estimation. We use the relation between the reference and each pixel for depth of each pixel. Finally, we obtain a volume of femur bone. The last step, we calculate the vBMD for osteoporosis diagnosis.

4.1 Contour Extraction

In system, a standard hip x-ray image shown in Fig. 4 is used as an input image to automatically extract femur contour. Bone contour extraction from x-ray image is a very important preprocessing step in computational analysis of medical imaging. The complexity is higher than that of CT and MR image segmentation because the regions delineated by bone contours are highly non-uniform in intensity and texture. Classical segmentation algorithms based on homogeneity are not applicable.

4.1.1 Modified Canny Edge Detection

Canny edge [17] detector takes gray scale image as an input to produce an output image showing position of the edges. It works as follows. The image is firstly smoothed by Gaussian convolution. Then, a simple 2D first derivative operator is applied to the smooth image to highlight regions of the image with high first derivative values. Using the gradient direction calculated, the algorithm performs non-maxima suppression to eliminate pixels whose gradient magnitude
is lower than its two neighbors along the gradient direction. The Canny edge detector [18] with a small smoothing effect is used to detect the boundary of femur head. Spurious edges with both low intensity value and low gradient magnitude values are removed. In brief, a pixel is marked as a non-edge point if 1) it is detected by Canny edge detector, 2) its intensity is lower than a threshold, and 3) its edge magnitude is lower than the same threshold. When the Canny edge detector is directly applied to the femur images, it will either produce too much noise, if a lower threshold is used, or lose some actual edges at the femoral head (Fig. 5). The idea is to preserve the edges at the femoral head and remove the noise at the same time by looking at the pixel intensity. Observation shows that the pixels on the bone region normally have higher intensity values than the noise.

4.1.2 Modified Active Contour Model with GVF

Active Contour Models (ACM) [19] are curves in an image domain which is movable under the influence of internal forces coming from within the curve itself and external forces computed from the image data. The internal and external forces are defined so that ACM will conform to the object boundary or other desired features within an image. The external force model that is closest in spirit to Gradient Vector Flow (GVF) is the distance potential forces. GVF uses the force balance as a starting point for designing a snake. We define a new static external force field called snake with the GVF field. The result of the modified Canny edge detector is shown in Fig. 6.

4.2 Finding Reference Point

First of all, the femoral head must be detected and its diameter is calculated. The construction of the level lines depends on the normal of the contour points. The centroid-radii model [20] samples a set of points from the outline of the object. A common approach is to use finite difference to estimate the derivative and hence derive the normal direction. To compute the normal of a contour point, neighborhood points around the point of interest is chosen. This set of points represents a segment of the contour. Given a set of points in 2D, two eigen-vectors and their eigenvalues are calculated. The eigenvector with the largest eigenvalues will point in the direction parallel to this segment of points and the other eigenvector gives the normal direction at the point of interest. Once the normal for each point on the contour has been calculated. After that the set of level lines can be computed. We must detect femoral head for finding femoral head diameter. Computing initial estimation of the level lines depends on the normal of the contour points and there are a few ways to compute the normal for a point on the contour. Let $P_1$ and $P_2$ denote the vector representation of two points on the contour and $N_1$ and $N_2$ be the unit normal. Then the line $L(P_1, P_2)$ that connects points $P_1$ and $P_2$ is a level line if

$$|N_1 \cdot N_2| \approx |(P_1 - P_2) \cdot n| \approx |(P_1 - P_2) \cdot N_2| \approx 1$$

In the current implementation, two orientations, $V_1$ and $V_2$ are similar i.e.

$$|V_1 - V_2| \geq 0.98$$

The orientation of the femur shaft can be computed by extracting the midpoints of the level lines on the shaft. Given level lines $(P_{1i}, P_{2i})$ and the midpoints $M_i = 1/2 (P_{1i} + P_{2i})$. First of all, we must detect femoral head for finding femoral head diameter. We get a couple points then we draw a level line. So these level lines are a diameter candidate of femoral head. Figure 7 shows that a midpoint of diameter is a reference point.

The concept of the reference point involves the pelvic bone region. Thus, the reference point can be divided into
two cases: the non-overlapping case and overlapping case. Figure 8 shows the concept of an overlap between midpoint and the pelvic bone region. Figure 8 (a) is a non-overlapping case that shows a value of a linear attenuation coefficient and the thickness of midpoint. We can use the values at the midpoint. Figure 8 (b) is the overlapping case. The midpoint is inside the pelvic bone region. The midpoint has three linear attenuation coefficients: the tissue, femoral head and pelvic bone. The midpoint is first determined. But the value of the midpoint cannot be related to the thickness of each pixel. Therefore, we must select another point not overlapped with the pelvic bone. Figure 9 (a) shows the 3D overlapped case. Figure 9 (b) shows the 2D plane for finding a new reference point value. The midpoint is not the reference point because it is not only the thickness of femur. It is the thickness of the femur bone and the pelvic bone. The midpoint is firstly determined as well. Therefore, we must move the midpoint to the new reference point that is moved within the femur head until outside the pelvic bone. We assume that the co-ordinate of new reference point of femur head is \((X, Y)\), while the co-ordinate of midpoint is \((0, 0)\). Therefore, we obtain intensity \(I_{(X,Y)}\) and the thickness \(t_{(X,Y)}\) at the new reference point.

\[
t_{(X,Y)} = 2R^2 = 2\sqrt{(X^2 + Y^2) - (R^2)}
\]

4.3 Estimation of Volume

This method is based on the fact that absorption of x-rays in any material, such as bone mineral and soft tissue like fat and muscle, is a function of the x-ray energy as well as the material itself. Using this fact, it is possible to separate the contribution to the x-ray absorption due to bone from that due to soft tissue if the transmitted x-ray intensity is measured at two different discrete energies. Thus we get a diameter of femur head that is assumed to be a sphere. We consider the reference point about the overlapped case. After we received the reference point of femur head then using Eq. (8), we can calculate the relative depth between two points. The Volumetric estimation is the last step of the entire process. All the reconstruction results can be viewed directly in three-dimensional and in an interactive way. User can change the surface geometry in three dimensions. Figure 10 shows the depth of each pixel.
required parameters are obtained from x-ray machine. To make a quantitative measurement of the bone mineral density of for instance the forearm it is not sufficient to just let a mono-energetic x-ray beam penetrate the hip and measure the transmitted intensity. The reason for this is that the x-rays will of course be attenuated by both bone and soft tissue (muscle and fat). Hence, it is impossible to separate the attenuation due to bone from that due to soft tissue. From the gradient vector flow, the energy function is:

\[
E = \int \int \left[ \mu (q_x^2 + q_y^2 + r_x^2 + r_y^2) \right] + |\nabla E|^2 \text{d}x \text{d}y
\]

The object is to minimize this energy function where \( E \) is an edge map \( E(x, y) \) obtained from the image, where \( \mu \) is a constant that is set according to the amount of noise present, and \( q_x, q_y, r_x, \) and \( r_y \) are the partial derivatives of \( q \) and \( r \) with respect to \( x \) and \( y \). \( E \) is the gradient vector normal to the edge \( E \) derived from the image. Using variational calculus, it can be shown that the GVF field can be computed by a pixel set corresponding to the minimum energy value is then illustrated as follows:

\[
C = \{(x_1, y_1), (x_2, y_2), (x_3, y_3)\}
\]

where \((x_i, y_i)\) is the pixel position in rectangular coordinates. Suppose the intensity of photon value \( I_p(x_i, y_i) \) is anti-proportional to the intensity of image pixel \( I(x_i, y_i) \)

\[
I_p(x_i, y_i) \propto \frac{1}{I(x_i, y_i)}
\]

Then, we get

\[
I_p(x_i, y_i)I(x_i, y_i) = K
\]

For the intensity equation, the photon intensity can be initially determined by

\[
I_p(x_i, y_i) = I_0 e^{-(\mu_s + \mu_b)t_b}
\]

Substitute \( I_p \) to Eq. (8).

\[
K = I(x_i, y_i)I_0 e^{-(\mu_s + \mu_b)t_b}
\]

Where \( I_0 \) is an incident intensity and \( t_b \) is the thickness of pixel \((x_i, y_i)\). The \( \mu_s \) is the linear attenuation coefficient of tissue, and \( \mu_b \) is the linear attenuation coefficient of bone. When compared the intensity between two pixels, we get

\[
\begin{align*}
I(x_1, y_1)I_0 e^{-(\mu_s + \mu_b)t_b} &= I(x_2, y_2)I_0 e^{-(\mu_s + \mu_b)t_b} \\
\ln I(x_1, y_1) - (\mu_s + \mu_b)t_b &= \ln I(x_2, y_2) - (\mu_s + \mu_b)t_b \\
\ln I(x_1, y_1) \ln I(x_2, y_2) &= (\mu_s + \mu_b)(t_b - t_b)
\end{align*}
\]

A term of \((t_b - t_b)\) is a difference of the thickness between two pixels. Since a femur head is assumed to be as sphere-like shape, a diameter of the head can be measured from the depth at the head center. From Eq. (8), \( I(x_0, y_0) \), an intensity of head center, is determined as the reference intensity. The Eq. (8) can be derived to

\[
\ln I(x_1, y_1) \ln I(x_0, y_0) = (\mu_s + \mu_b)(t_b - t_b)
\]

where \( I(x_i, y_i) \) is the intensity of the considered pixel. The thickness of the considered pixel \((x_i, y_i)\) is derived from Eq. (10).

\[
t_b = t_b(x_1, y_1) = \ln I(x_1, y_1) - \ln I(x_0, y_0) + t_b
\]

where \( \mu = \mu_s + \mu_b \)

Give the small volume of femur with \( dV \). This volume can be found by the multiplying the small are of cross section of femur \((d\Phi)\) with its depth (thickness) as the following equation

\[
dV = t_b(\phi)d(\phi)
\]

\[
V = \int t_b(\phi)d(\phi)
\]

\[
V = \int \int t_b(x, y)d(x, y)
\]

However, the image is discrete. The double integral can be replaced by the summation.

\[
V = \sum t_b(x_i, y_i)\Delta\Phi_i = \sum t_b(x_i, y_i)\Delta\Phi_i
\]

Where \( i \) is on index of pixel in pixel set \( C \) and

\[
\Delta\Phi_i = \sigma; \forall_i
\]

\( \sigma \) is called the constant of a small cross section area. After that,

\[
V = \sum t_b(x_i, y_i)\sigma
\]
Table 1  The result measure of the geometrical objects (n = 24) compared between object density and computed density.

<table>
<thead>
<tr>
<th>Mass (g)</th>
<th>areal object (DEXA) (g/cm²)</th>
<th>Object Density (10^2 g/cm²)</th>
<th>Computed Density (10^2 g/cm²)</th>
<th>Density Difference (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>19.45</td>
<td>0.92 (0.01)</td>
<td>30.35 (1.55)</td>
<td>30.1 (0.02)</td>
<td>0.82%</td>
</tr>
</tbody>
</table>

Table 2 Characteristic of experiment subject.

<table>
<thead>
<tr>
<th>Variables</th>
<th>Men (n = 120)</th>
<th>Women (n = 180)</th>
<th>Mean difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (years)</td>
<td>50.3 ± 18.0</td>
<td>52.7 ± 16.2</td>
<td>−2.4</td>
</tr>
<tr>
<td>Weight (Kg)</td>
<td>59.1 ± 7.2</td>
<td>55.2 ± 11.3</td>
<td>3.8</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>162.2 ± 6.1</td>
<td>153.3 ± 7.4</td>
<td>8.9</td>
</tr>
<tr>
<td>Body mass index (Kg/m²)</td>
<td>21.9 ± 2.2</td>
<td>25.2 ± 4.3</td>
<td>−4.7</td>
</tr>
<tr>
<td>Cross-section area (cm²)</td>
<td>9.5 ± 1.6</td>
<td>7.1 ± 1.3</td>
<td>2.4</td>
</tr>
</tbody>
</table>

Substitute \( tb(x_i, y_i) \) from Eq. (14). Then,

\[
V = \sum \left( \frac{\ln I(x_i, y_i) - \ln I(x_0, y_0)}{\mu} + tb_0 \right) \sigma \tag{18}
\]

\[
V = \sum \ln I(x_i, y_i) \tag{19}
\]

Where \( \eta \) is volume reference coefficient and can be computed by

\[
\eta = n\sigma \left( tb_0 - \frac{\ln I(x_0, y_0)}{\mu} \right) \tag{20}
\]

4.4 Estimation of Volumetric BMD

Body weight was measured using an electronic balance scale (accurate to the nearest 0.1 kg) and standing height (accurate to the nearest 0.1 cm). Body mass index (BMI) was calculated as the ratio of weight (kg) over height squared \((m^2)\). The areal Femoral Bone Mineral Density (aFBMD), unit \((g/cm^2)\), was measured by DXA using the Hologic QDR4000 system, giving BMD results \((g/cm^3)\). Statistical analyses were performed using SPSS version 11. Data analysis was performed separately for men and women. Descriptive results were expressed as means, standard deviations (SD) and per cent. Comparisons between men and women were made using the unpaired t-test. The results are classified into two groups. The first group is that of the experiments with a geometry shape such as a sphere, cylinder and cubic for determination a volume. The second group of experiments is a BMD of x-ray image \((g/cm^3)\) compare with that of a BMD DEXA \((g/cm^2)\), using a BMC value of DEXA. The experiments were done on 300 images, those of 180 females and 120 males, aged 20 years to 80 years old. The size of testing images was 295×350 mm². The settings of the X-ray unit were 100 kV, 100 mA, 2 sec exposure time. Radiographs were developed with Kodak chemicals. A simple model femur is used by the algorithm to extract the femur contours in the test images. The error of an extracted contour is measured in terms of the mean error between the points on the extracted contour and their corresponding points on the manually marked contour. Also, the correlations between the BMD of the input x-ray image and CT-scanned images is computed.

The first group, examples of the results of estimating the density in Table 1 is to find the volume of several geometric shapes. A set of 24 images were obtained. Of these 24 images, 8 images contained a sphere shape, 8 images contained a cube shape, and 8 images contained a cylinder shape. Error of the density is quite low because it is easy to find the reference point so that the calculated volume is computational accurate compared to the real volume. The result shows 0.8% density difference between a real density and a computed density.

Table 2 shows the results of the second experiment that is total of 120 men and 180 women with complete data analysis. The average age of the men and women was 50.3 ± 18 and 52.7 ± 16.2 years, respectively. The average age of men was 50 years. Men were significantly heavier and taller, but had lower BMIs than women. The bone parameters, Cross-Section Area (CSA), were significantly higher in men than in women (p < 0.001). Figure 11 shows the relation between sex, age and bone mineral density. It presents aBMD (volume g/cm³).
and vBMD. Table 3 is the comparison between the volume of femur measuring from CT-scan and that obtained using our method.

6. Discussion

The authors found that men had significantly higher FBMC, CSA and vBMD than women; however, the difference in aFBMD between men and women, using the usual aFBMD measure, disappeared when an estimate of vFBMD was used. The results presented in this paper demonstrate that a bone mineral density of the femur can be obtained from a single calibrated x-ray. The actual size, and hence the volume, of the bone remains unchanged, but the cortical regions get thinner and the bone shows osteoporosis. There is less specific bone tissue in relation to bone marrow and the additional space created is filled with fat. As far as the chemical composition is concerned osteoporosis bone is not distinguishable from normal and the relationship between organic matter and mineral is unchanged. One approach to this problem is to embed the examined body part in some material that resembles muscle and other types of soft tissue in terms of x-ray attenuation, for instance water. In this way the thickness of soft tissue and soft tissue equivalent material surrounding the bone is constant. The absorption profile will then show a fixed base line due only to absorption in soft tissue and the surrounding material (water). But our method can compute a real depth. Our method can compute the object depth even though it is not a real geometric shape as we can find the reference point compare to all points in the image.

7. Conclusion

This paper proposes a noninvasive bone volume measurement method from a single x-ray image using x-ray atten-
uation on radiography and medical knowledge. This paper has demonstrated how the bone mineral density can be estimated in three dimensions even when only a single x-ray image is available. The medical knowledge is important because a geometrical shape can inform other dimensions. This research focused on the bone part whose shape we know. We can estimate the volumetric of the region of interest. This technical method shows the experimental results that show 3.23% difference in volume compared to the BMD of CT-SCAN, which are in good agreement. This research anticipate that useful information from the medical knowledge is improved, the automated estimation BMD of x-ray image can be obtained.

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