Relationship between resolution and partial volume effect among µCT, MDCT and SDCT

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Abstract
Partial volume effect is defined as the loss of accuracy for small objects caused by the low resolution of an imaging system. With low resolution computed tomography (CT), the trabecular bone and cavity are mixed and the brightness representing each of the spaces is averaged. Therefore, information regarding bony microstructure is absent. In this study, the partial volume effect was evaluated for multi-detector row CT (MDCT) and single-detector row CT (SDCT) in comparison with micro CT (µCT). Obvious and typical geometric patterns of healthy and osteoporotic bones were used to create virtual sectional images of various resolutions. Six parameters were evaluated: areal bone mineral density (aBMD), volumetric BMD (vBMD), bone volume (BV), bone mineral content (BMC), frequency distribution density of BMD (FDD) in the image, and the orientation angle of the bone. vBMD and BV values were dependent on the CT resolution, whereas aBMD and BMC values showed constant values independent of the resolution. Therefore, aBMD and BMC do not require high resolution CT and could be useful for clinically evaluating trabecular bone volume. Regarding FDD, the number of pixels with intermediate brightness increased as CT resolution decreased, and FDD converged on specific brightness representing aBMD. In addition, µCT estimated the bone orientation angle correctly, MDCT estimated the correct angle only for osteoporotic images, and SDCT was unable to estimate the angle. Many more cavities were present in the osteoporotic model than the Healthy model and the distribution of bone was sparse, which could have decreased the partial volume effect and enabled the major orientation angle of the bone to be distinguished. These findings suggest that MDCT could be useful for the clinical evaluation of osteoporotic bone structure.

Keywords: Micro bone structure, Multi-detector row computed tomography (MDCT), µCT, Partial volume effect, Bone mineral density (BMD)

1. Introduction

The evaluation and analysis of bone is dependent on the CT resolution, and is especially important in the clinical setting. CT devices can be broadly classified into three categories according to their resolution: micro CT (µCT), multi detector-row CT (MDCT), and single detector-row CT (SDCT). µCT has high resolution, in the range 1–100 µm, but has the drawbacks of high radiation exposure and poor temporal resolution (Buie et al., 2007; Cooper et al., 2007; Tsubota et al., 2011) and is therefore not suitable for scanning patients. MDCT has resolution in the range 100–500 µm (Bauer et al., 2014) and exposure dose less than that of µCT, and is used for clinical diagnosis. SDCT has lower resolution (>500 µm; Duan et al., 2019), lower exposure dose, and shorter imaging time compared with µCT and MDCT, and is widely
used clinically (Duan et al., 2019).

To evaluate the microstructure of trabecular bone, it is necessary to consider the relationship between CT resolution and the diameter of trabecular bone, which is ~100 µm (Best et al., 2017; Boutroy et al., 2005). The high resolution of µCT enables evaluation of trabecular structure and direct measurement of trabecular orientation and volumetric bone mineral density (vBMD) (Burghardt et al., 2010; Boutroy et al., 2005; Malekzadeh et al., 2019). In contrast, as the resolution of MDCT is less than the trabecular diameter, one MDCT voxel includes both bone and cavity. When the CT value of a voxel expresses the average of bone as well as cavity, this phenomenon is termed partial volume effect. MDCT cannot adequately measure the structure of trabecular bone; however, MDCT has shown significant correlation with µCT in terms of areal bone mineral density (aBMD) (Bissinger et al., 2010; 2016), trabecular thickness (Gregorio et al., 2015; Malekzadeh et al., 2019), and bone density (Bauer et al., 2014). Accordingly, MDCT preserves the trabecular structure qualitatively. In contrast, because the resolution of SDCT is approximately five times the trabecular diameter, SDCT is inadequate for evaluating trabecular structure.

MDCT can be used clinically to evaluate bone quality parameters such as microstructure in cancellous bone, but the partial volume effect in MDCT relative to µCT is insufficiently understood in quantitative terms. Quantitative elucidation of characteristics of MDCT could be helpful to measure amount of the bone contents and to structure of the trabecular bone even in the spine or the femoral neck in the deep body. Also, MDCT should be useful to diagnose osteoporosis or evaluate drug treatment. Therefore, the objective of this study was to elucidate the relationship between trabecular structure and CT resolution by evaluating partial volume effect in virtual sectional images of geometrical patterns similar to those of typical trabecular bone. Bone structural parameters related to BMD and the major orientation angle of trabecular bone were calculated and compared for virtual sectional images of three different CT resolutions, and the characteristics were evaluated.

2. Material and Methods

2.1 Outline

We prepared two virtual trabecular models for comparison among µCT, MDCT, and SDCT: a healthy bone model and an osteoporotic bone model. To evaluate the characteristics of the three kinds of CT theoretically, we used the virtual trabecular models as ideal conditions. Each bone model was rotated 0°, 15°, and 30° to represent the major orientation angle of trabecular bone, and six planar models were constructed. Three image types were created for each model, with resolution according to the pixel size of µCT, MDCT, and SDCT. In total, 18 cross-sectional images were created. In addition, to confirm the SDCT characteristics in actual patients, we reconstructed SDCT images from MDCT images of the L1 vertebra obtained from one patient. In 20 images, we converted the brightness of each pixel into bone mineral density (BMD (g/cm³)) using the equations proposed by Keyak et al. (1998). The BMD values of each pixel were used to calculate five bone structural parameters related to BMD. In each of the 20 cross-sectional images that included the patient CT data, the major orientation angle in trabecular bone was estimated using the gradient tensor of brightness (Tsubota et al., 2002). The five bone structural parameters and the major orientation angle are calculated regardless of CT resolution, and do not need specific device. Therefore, total of six parameters should be adequate to compare among µCT, MDCT and SDCT. To elucidate the MDCT characteristics, we compared the bone mineral parameters related to BMD and the orientation angle among µCT, MDCT, and SDCT.

The Ethics Committee of our institute approved the use of patient data in this study and written informed consent was obtained from the patient prior to scanning.

2.2 Virtual trabecular models and clinical CT images

We scanned virtual trabecular models of cancellous bone with constant orientation. As shown in Fig. 1 (a), the virtual trabecular model has a vertical column as the major trabecular orientation and a horizontal bridge beam as the minor orientation. In the healthy structure (Healthy model, Fig. 1 (a)), the diameter of the vertical column was 100 µm and the diameter of the horizontal bridge was 60 µm. In the osteoporotic structure (Osteoporotic model, Fig. 1 (a)), the diameter of the vertical column was 160 µm, and the diameter of the horizontal bridge was 60 µm based on the clinical data reported by Buie et al. (2007). The healthy and osteoporotic structures were aligned to make a 4500 × 4500 µm image representing the trabecular structure with an orientation angle of 0°. The models were also rotated by each of 15° and 30° to produce trabecular structure models with orientation angles of 0°, 15°, and 30°. We obtained virtual µCT, MDCT, and SDCT cross-sectional images of these six models with pixel sizes of 10, 300, and 500 µm, respectively. The area
averaging method (Tian et al., 2012) was used to convert the cross-sectional image into an equivalent resolution for each CT type. We assumed a slice thickness of 10 µm for all CT types as a common condition, and compared the bone structural parameters in terms of the planar difference in resolution. We prepared a total of 18 cross-sectional images, as shown in Fig. 1 (b).

![Trabecular models and virtual cross-sectional images](image)

From the patient MDCT images of the L1 vertebra (Fig. 2(a)) we extracted images of size 4680 × 4500 µm (Fig. 2(b)), similar to those of the virtual cross-sectional images. The patient MDCT images were acquired as an ordinary image diagnosis. Also, CT device was Aquilion PRIM 80 (Canon Medical System Corporation, Ptawara, Japan), and it took about 10 seconds to acquire one image series. Table 1 lists the patient and scan conditions. The MDCT images were converted to low-resolution images equivalent to SDCT by the method used to make virtual SDCT images. Based on the actual clinical images, bone structure parameters related to BMD were compared between MDCT and SDCT. The µCT parameters were not obtained because only MDCT was obtained in the patient.

2.3 Conversion of CT image to BMD

In the 18 virtual sectional images and 2 patient images, pixel brightness was gradated from 0 (black) to 255 (white). The gradated values were assumed to be the spatial or volumetric ratio of cavity and bone. Graded brightness was translated to Young’s modulus and then to BMD. Bone with a pixel brightness value of 255 was assumed to be equivalent to cortical bone, as 20,000 MPa; whereas a cavity with a value of 0 was assumed as 0.001 MPa to avoid discontinuity of Young’s modulus (Keyak et al., 1998). We assumed that Young’s modulus was proportional to pixel brightness. Young’s modulus was translated linearly to BMD using the equations proposed by Keyak (1998).
2.4 Bone structural parameters related to BMD

Four bone structural parameters related to BMD were calculated. Areal BMD (aBMD) is defined as BMD for a unit image, calculated as

\[ aBMD = \frac{1}{N_{\text{vox}}} \sum_{i=1}^{N_{\text{vox}}} \rho_i \]  

where \( \rho \) is the BMD value of the \( i \)th voxel and \( N_{\text{vox}} \) is the total number of voxels in the image.

Bone volume (BV) cm\(^3\) is the estimated bone volume in a specific image, calculated with voxels of BMD >0.0001 g/cm\(^3\) as bone because the BMD value of cavity is \( \leq 0.0001 \) g/cm\(^3\).

Volumetric BMD (vBMD, g/cm\(^3\)), BMD per unit volume, is calculated as

\[ vBMD = \frac{1}{BV} \sum_{i=1}^{N_{\text{vox}}} \rho_i \quad (\rho_i > 0.0001 \text{[g/cm}^3\text{]}) \]  

Bone mineral content (BMC, g), the weight of bone mineral, is calculated as the product of BV and vBMD.

2.5 Frequency distribution density of BMD

As shown in Fig. 2, there are gradations in brightness in a CT image that correspond to gradated BMD values scattered throughout the image. As the BMD values and their frequencies in an image are dependent on the trabecular structure, the frequency distribution density of BMD (FDD) can be determined. As well as a histogram of BMD, FDD allows us to evaluate a distribution of BMD with FDD. We obtained FDD in the three types of CT images and in the patient CT images after normalizing by the total number of voxels in the images. FDD was compared among \( \mu \)CT, MDCT, and SDCT to evaluate the relationship between CT image resolution and partial volume effect.

2.6 Orientation of trabecular bone on CT images

We estimated the orientation angle of trabecular bone using the brightness gradient tensor shown in Eqs (3) to (7) (Bathursta and Rothenburg, 1990). The difference in brightness and the distance from the target voxel to the surrounding voxels were calculated using Eqs (3) and (4), respectively, as
\[
\Delta L_j = L_j - L_t \tag{3}
\]
\[
\Delta d_j = |\mathbf{x}_j - \mathbf{x}_t| \quad (\Delta d_j < r) \tag{4}
\]

where \( L_t \) is the target voxel, \( L_j \) is the brightness of the \( j \)th voxel around the \( L_t \) voxel, and \( \Delta L \) is the difference between \( L_t \) and \( L_j \). Also, \( \mathbf{x} \) is the position vector of the target voxel, \( \mathbf{x}_j \) is the position vector of the \( j \)th voxel around \( \mathbf{x} \), and \( \Delta d_j \) is the distance from \( \mathbf{x} \) to \( \mathbf{x}_j \). Here, variable \( j \) is related to the radius of detection \( r \) shown lately, and the \( j \)th voxel is in the range of the radius \( r \).

The weight function \( w_j \) is determined according to Eq. (5). \( w_j \) has a heavy weight for a near voxel and a light weight for a far voxel, depending on the distance. \( r \) is the radius of detection (µm). Brightness gradient tensor \( \mathbf{F} \) is calculated using \( w_j \), as in Eq. (6).

\[
w_j = \left( 1 - \frac{\Delta d_j}{r} \right)
\]

\[
\mathbf{F} = \sum \left\{ \Delta L_j \cdot w_j \cdot (\mathbf{n}_j \otimes \mathbf{n}_j) \right\}
\]

Here, \( \mathbf{n} \) is the unit vector from \( \mathbf{x} \) to \( \mathbf{x}_j \). Thus, the radius \( r \) affects the sensitivity for detecting the major orientation angle of trabecular bone. As shown later, we decided the smallest radius \( r \) using images rotated by 30°.

Because the brightness gradient tensor \( \mathbf{F} \) is a positive definite symmetric matrix, it is uniquely decomposed into an orthogonal matrix \( \mathbf{R} \) and a diagonal matrix \( \lambda \) with eigenvalues, as follows

\[
\mathbf{F} = \mathbf{R} \lambda \mathbf{R}^T
\]

When we set the eigenvector \( \mathbf{e}_\text{min} \) concerning the minimum eigenvalue \( \lambda_{\text{min}} \), \( \mathbf{e}_\text{min} \) is the direction vector of the smoothest gradient of brightness, and continuity of brightness; i.e., the orientation of trabecular bone or cavity. Thus, we calculated the angle \( \theta \) between \( \mathbf{e}_\text{min} \) and the vertical direction as the principle orientation angle at a target pixel. From any \( \theta \) on a CT image, we calculated a normalized FDD concerning \( \theta \), and then assumed that the direction angle to the peak frequency was the major orientation angle \( \theta \) of trabecular bone on the CT image.

3. Results

3.1 Bone structural parameters of µCT, MDCT, and SDCT

Figure 3 shows aBMD in the Healthy and Osteoporotic models and in the patient data. aBMD values are shown at orientation angles of 0°, 15°, and 30° in the Healthy and Osteoporotic models. There was little variation in aBMD value (<0.0003 g/cm³) among the CT types and orientation angles in the Healthy model (Figs 3(a)). There was also little variation in aBMD in the Osteoporotic model (Fig. 3(b)), with aBMD values slightly higher for an orientation angle of 0° than for 15° and 30°. aBMD values were lower in the Osteoporotic model than the Healthy model. The standard deviation (S.D.) of BMD differed among the CT types (Fig. 3(a, b)), and increased with CT resolution. S.D. of BMD was greater for higher resolution CT because µCT has only two values of pixel brightness, 255 and 0 for bone and cavity.
respectively. However, MDCT and SDCT have intermediate values due to the partial volume effect. As shown in Fig. 3 (c), aBMD values and S.D. were similar for MDCT and SDCT in the patient data.

Figure 4 shows vBMD in the Healthy and Osteoporotic models and in the patient data. In both of the virtual models, there was no change in vBMD for $\mu$CT among the orientation angles. vBMD was much lower for MDCT and SDCT than for $\mu$CT. In the Healthy model (Fig. 4(a)), the greatest difference in vBMD between MDCT and SDCT was 0.0814 g/cm$^3$ at an orientation angle of 0°, and the difference was a little larger in MDCT than SDCT at angles of 15° and 30°. In the Osteoporotic model (Fig. 4(b)), the difference between MDCT and SDCT decreased with increasing orientation angle. There was no difference in vBMD values in the patient data (Fig. 4 (c)).

Figure 5 shows the BV in the Healthy and Osteoporotic models and in the patient data. The BV value for $\mu$CT showed little change in both models regardless of angle, but was slightly higher in the Healthy model. This result is expected because BV differs between healthy and osteoporotic bone. In the models, BV for MDCT increased with increasing orientation angle, and the BV values were much higher for MDCT and SDCT than for $\mu$CT. This finding indicates pseudo BV for MDCT and SDCT due to partial volume effect. There was little difference in BV between MDCT and SDCT in the patient data (Fig. 5(c)).

Figure 6 shows BMC in the Healthy and Osteoporotic models and in the patient data. BMC is constant among CT types for each orientation angle. In the Healthy model (Fig. 6 (a)), BMC is slightly less at 0° than at the other angles. In the Osteoporotic model (Fig. 6 (b)), BMC is higher at 0° than at the other angles, and BMC values are almost the same between 15° and 30°. For both models, BMC values are the same at each orientation angle among $\mu$CT, MDCT, and SDCT. There was no difference in BMC between MDCT and SDCT in the patient data (Fig. 6(c)).
3.2 FDD of BMD

Figure 7 shows FDD in the Healthy and Osteoporotic models and in the patient data. Both of the models were evaluated at an orientation angle of 0°. In the Healthy model, BMD is constant for SDCT and corresponds to aBMD, and there are four values of BMD for MDCT, whereas there are only two BMD values for µCT (for each of bone and cavity). In the Osteoporotic model, similar to the Healthy model there are only two BMD values for µCT, but the FDD values are different to the Healthy model because the Osteoporotic model has fewer bony voxels compared with the Healthy model. There are four values for MDCT, similar to the Healthy model, and SDCT has many gradated values of BMD. In the patient data, the FDD of MDCT is similar to SDCT, and the distribution of BMD shows gradual change.

3.3 Relationship between orientation angle and radius of detection

Figure 8 shows the FDD for angles relative to the estimated angle for various radii of detection $r$ in the images for an orientation angle of 30°. In the µCT images of the Healthy and Osteoporotic models, the orientation angle does not align with 30° for a radius of 15 µm (Fig. 8 (a) and (b)). For radii of detection of 105–705 µm, peak density occurs around 30°; in particular, peak frequency distribution density values occur at an estimated angle of 30.1° for a radius of 705 µm in the Healthy model and a radius of 605 µm in the Osteoporotic model. The estimated angles for MDCT are shown in

Fig. 7 FDD in Healthy and Osteoporotic models at 0°, and patient data. FDD is expressed for cross-sectional images.
Table 2  Optimal radius $r$ and estimated angle for the orientation angle of 30° in the Healthy and Osteoporotic models

|                | µCT | MDCT | SDCT |
|----------------|-----|------|------|
| Healthy        |     |      |      |
| radius (µm)    | 705 | 1950 | 2750 |
| angle (°)      | 29.8| 1.55 | -76.0|
| Osteoporotic   |     |      |      |
| radius (µm)    | 605 | 750  | 1750 |
| angle (°)      | 30.1| 27.1 | 6.35 |

Fig. 8 Relationship between FDD of estimated orientation angle and radius of detection $r$ for an orientation angle of 30°.
Fig. 9 Relationship between FDD of orientation angle and estimated orientation angle.
The orientation angle is not estimated correctly for the Healthy model. For the Osteoporotic model, however, the peak value is close to 30° for all radii except 450 µm; in particular, the estimated angle is 27.1° for a radius of 750 µm. In the SDCT images, there is no obvious peak density in the Healthy model (Fig. 8 (e)), and there is no obvious peak for a radius of 2750 µm. In the Osteoporotic model, there is an obvious peak at 6.4° for a radius of 1750 µm. The orientation angle is estimated incorrectly because there are insufficient radii to detect the orientation angle.

The optimal values for the three types of CT in both models are summarized in Table 2. The same radii were used for MDCT and for the patient CT data.

3.4 Estimation of Orientation angle

We estimated the orientation angle in the CT images using the radii shown in Table 2. Figure 9 shows the frequency distribution density in all CT types for various orientation angles. In images with an orientation angle of 0° (Fig. 9 (a) and (b)), the angle of 0° was estimated correctly by MDCT and SDCT. The angle could not be estimated for SDCT images, which have only one grey color and no gradient of brightness. In the Osteoporotic model, the angle of 0° was estimated correctly in all CT image types. In images with an orientation angle of 15° (Figure 9 (c) and (d)), only µCT estimated the orientation angle correctly (15° in the Healthy model and 15.1° in the Osteoporotic model). MDCT showed peak FDD at –17.5° and a secondary peak at 17.0° in the Healthy model (Fig. 9(c)), and at 11.1° in the Osteoporotic model (Fig. 9(d)). In images with an orientation angle of 30°, µCT estimation was accurate in both models. MDCT estimated the orientation angle as 27.1° in the Osteoporotic model, but estimated the angle incorrectly in the Healthy model. In the patient CT data shown in Fig. 9(g), the orientation angle was estimated as 4.5° by MDCT and as 2.9° by SDCT.

4. Discussion

4.1 Evaluation of bone structural parameters (aBMD, vBMD, BV, and BMC)

Although the definitions of aBMD and vBMD are similar, aBMD is related to the total number of voxels as an area, whereas vBMD is related to actual volume in a CT image. Therefore, vBMD is more sensitive to the partial volume effect in bone structures. Comparing Figs 3 and 4, the vBMD values show a clear difference in sensitivity among the CT types, and are particularly high for µCT. In addition, as µCT can detect the border between trabecular bone and cavity, the vBMD values for µCT are constant regardless of orientation angle. Therefore, as resolution increases, the value of vBMD increases. The vBMD values were slightly higher for MDCT than for SDCT; however, there was no difference between MDCT and SDCT in the patient data (Fig. 4(c)).

The BV values estimated by SDCT are large (Fig. 5), for the reason that the number of voxels with low or middle brightness can increase with low-resolution CT because the geometric volumes of trabecular bone and cavity are averaged by the partial volume effect. Accordingly, low-resolution CT can cause pseudo enhancement of BV. Additionally, information regarding the orientation angle of trabecular bone can be lost with low-resolution CT because of the homogeneity of brightness on the images. In the patient CT data (Fig. 5 (c)), BV values are slightly higher for SDCT than for MDCT, similar to the results reported for other virtual models.

Regarding BMC, the values were the same for µCT, MDCT, and SDCT for each angle, for both models (Fig. 6). As BMC is the product of vBMD (as available BMD related to volume) and BV (as available volume), the BMC value indicates the actual bone mineral content. vBMD (Fig. 4) and BV (Fig. 5) are both affected by CT resolution, and their values differ among the CT types. However, BMC (Fig. 6) appears to be less affected by CT type. Therefore, BMC could be useful for evaluating the amount of bone itself. There is no difference in BMC between MDCT and SDCT in the patient data. In the Healthy model, the BMC value increases slightly with increasing trabecular angle, whereas in the Osteoporotic model, the BMC value decreases slightly as the angle increases. The dependency of the inclination is similar to that for aBMD, but may be enhanced for BMC because BMC is the product of vBMD and BV. Theoretically, BV might be estimated more correctly when we consider not only the BMD value on a target voxel, but also BMD values of voxels around a target voxel. This is because BMD values depend on the brightness of a target voxel in this study, and the brightness values are averaged with or distributed to around voxels by partial volume effect. However, our results indicate that BMC should include the changes by partial volume effect. Therefore, BMC might be suitable for evaluating the amount of bone independent of the CT resolution.

Our results for aBMD, vBMD, BV, and BMC agree with those published previously for actual CT images by Bissinger et al. (2016). The present patient data showed similarity with our virtual results. Therefore, our virtual CT
images could be considered equivalent to genuine CT images, which suggests that our method for calculating bone structural parameters could be feasible.

4.2 Frequency distribution density of BMD (FDD)

Because mean image brightness is used in aBMD, vBMD, VB, and BMC, these values do not represent the frequency distribution of brightness in the images. In the present study, we used FDD as a measure of the frequency distribution. Comparison of FDD among µCT, MDCT, and SDCT is useful for evaluating change in BMD due to the partial volume effect. At the lowest resolution, BMD converges to a single value, and we assumed the converged BMD value to be aBMD. Thus, even though aBMD is a single value, the FDD comprises all aBMD values.

As shown in Figure 7(a) and (b), the frequency of intermediate brightness values increased when resolution was decreased, due to partial volume effect. µCT has only two BMD values, bone and cavity, MDCT has four values, and SDCT has one intermediate value for the Healthy model and several values for the Osteoporotic model. As expected, the FDD of SDCT and aBMD were identical in the Healthy model, and in the Osteoporotic model, the FDD of SDCT was close to that of aBMD. Concerning the crossing points between FDD and aBMD, µCT intersected aBMD at a high FDD value, and MDCT intersected aBMD beneath this value in both models. Among the CT types, SDCT intersected aBMD at the lowest FDD value, which indicates that the cross point between the FDD of MDCT and aBMD could be useful for estimating the frequency distribution of bone or cavity as porosity. As FDD of µCT intersects aBMD at a higher point than does MDCT, we were able to estimate FDD of trabecular bone because FDD of µCT has only two values, bone and cavity.

In evaluation of the patient data, we estimated FDD using the MDCT and simulated SDCT images. In the patient CT images, FDD was distributed continuously but not discretely. Figure 7 (c) shows that MDCT and aBMD intersected at approximately 73%, similar to Fig. 7(b). Therefore, it could be expected that porosity in the patient is at least 73%.

4.3 Relationship between orientation angle and radius of detection r

As shown in Fig. 8, the estimated angle was about 30° for µCT in the Healthy model when the radius of detection was 105–705µm. Because the detection diameter was smaller than the size of the cavity, the contour of cancellous bone could be identified by the brightness gradient tensor, and the angle could then be estimated. Also, the accuracy of angle estimation was highest at radii of 605 and 705 µm. When the detection radius is large, a higher number of evaluation pixels are required, which can improve the accuracy of the estimation.

MDCT was unable to estimate the orientation angle in the Healthy model; whereas in the osteoporosis model, MDCT estimated the angle within 2.9° for a detection radius of 750 µm. This finding is due to the relationships among trabecular size, cavity size, and CT resolution. In the Healthy model, the minimum distance between trabecular bone and cavity was 400 µm, whereas the minimum radius of MDCT was 450 µm. In this case, the radius r should be less than the microstructure of the bone. In the Osteoporotic model, the minimum length of trabeculae and cavities is 740 µm, which is longer than the minimum MDCT radius. In addition, SDCT was unable to estimate the orientation angle. The detection radius of SDCT was larger than the trabecular structure size in the Healthy model and the Osteoporotic model.

As mentioned above, estimation of the orientation angle using the brightness gradient tensor requires the detection radius to be less than the size of the trabecular bone or cavity.

4.4 Estimation of orientation angle in trabecular bone

As shown in Fig. 9, µCT could estimate all orientation angles in both models. In contrast, SDCT could not estimate the orientation angle (Fig. 9 (a)), because SDCT images are a uniform grey color. The distribution of brightness disappeared due to the partial volume effect.

MDCT could not estimate the orientation angle at the first maximum peak of FDD in the Healthy model (Fig. 9(c)) and (e). As shown for MDCT in Fig. 1 (b) (Healthy), the partial volume effect increased the intermediate brightness values and reduced obvious patterns. Therefore, the original orientation angle was estimated correctly by the second peak rather than the first maximum peak in Fig. 9 (c), and similarly in Fig. 9 (e).

SDCT was also unable to estimate the correct orientation angle. In MDCT and SDCT, the obvious brightness patterns disappear due to the partial volume effect, as shown in Fig. 1 (b), which can cause misestimation of orientation angle. However, in the osteoporosis models shown in Fig. 9 (d) and 9 (f), MDCT could estimate the orientation angle because the detected radius r was shorter than the shortest length of trabecular bone or cavity. The images of the Osteoporotic
model retain their trabecular pattern more clearly than those of the Healthy model because the distribution of bone pixels in the Osteoporotic model is insufficiently dense to mix with cavity pixels in the partial volume effect. Accordingly, it is difficult for MDCT to estimate the orientation angle of trabecular bone in dense trabecular bone, but MDCT could be suitable for clinical evaluation of thin (osteoporotic) trabecular structure.

In the patient data as shown in Fig. 9 (g), we estimated an angle of 4.5° with MDCT and 2.9° with SDCT, which is a difference of 1.6°. The FDD of MDCT was less than that of SDCT, but we did not validate the comparison of FDD among the different CT types; in any case, Fig. 9 (d) and (f) showed that MDCT estimated the angle correctly. Therefore, the major orientation of trabecular bone could be 4.5°.

4.5 Limitations

Because we used patient MDCT images rather than μCT images as the real data, we could not confirm the orientation angle of trabecular bone or porosity at the μCT level, and our analysis was therefore not validated. To avoid the excessive exposure dose to patients that is unavoidable with μCT, cadavers are generally used to compare the three types of CT images evaluated here (Ito et al., 2005). Artificial bones can also be used for this purpose (Liu et al., 2012), and high-resolution peripheral quantitative CT (HR-pQCT) (Burghardt et al., 2007). In contrast, in the present study we used virtual CT images, which are approximation of the actual structure of the trabecular bone. Also, the images do not have noises and ideal condition is assumed in this study. However, the resulting images were accurate and quantitative data was obtained. Moreover, our method allows us to evaluate the characteristics of CT theoretically. Hence, our method could be validated, and it was adequate for evaluating the characteristics of BMD parameters for various levels of CT resolution.

5. Conclusion

The present study found that for μCT, MDCT, and SDCT, the partial volume effect had no effect on aBMD or BMC, regardless of resolution. In contrast, vBMD, BV , the orientation angle of the trabecular bone, and FDD depended on the distribution of brightness, which is affected by the partial volume effect. In addition, estimation of the trabecular bone angle was more accurate with μCT than with MDCT and SDCT. In the case of osteoporotic patients, however, MDCT appears to be suitable for evaluating the orientation angle in the clinical setting.

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