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FULL PAPER

Accuracy of the raw-data-based effective atomic numbers and monochromatic CT numbers for contrast medium with a dual-energy CT technique

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Objective: To evaluate the accuracy of raw-data-based effective atomic number (Z_{eff}) values and monochromatic CT numbers for contrast material of varying iodine concentrations, obtained using dual-energy CT.

Methods: We used a tissue characterization phantom and varying concentrations of iodinated contrast medium. A comparison between the theoretical values of Z_{eff} and that provided by the manufacturer was performed. The measured and theoretical monochromatic CT numbers at 40–130 keV were compared.

Results: The average difference between the Z_{eff} values of lung (inhale) inserts in the tissue characterization phantom was 81.3% and the average Z_{eff} difference was within 8.4%. The average difference between the Z_{eff} values of the varying concentrations of iodinated contrast medium was within 11.2%. For the varying concentrations of iodinated contrast medium, the differences between

the measured and theoretical monochromatic CT values increased with decreasing monochromatic energy. The Z_{eff} and monochromatic CT numbers in the tissue characterization phantom were reasonably accurate.

Conclusion: The accuracy of the raw-data-based Z_{eff} values was higher than that of image-based Z_{eff} values in the tissue-equivalent phantom. The accuracy of Z_{eff} values in the contrast medium was in good agreement within the maximum SD found in the iodine concentration range of clinical dynamic CT imaging. Moreover, the optimum monochromatic energy for human tissue and iodinated contrast medium was found to be 70 keV.

Advances in knowledge: The accuracy of the Z_{eff} values and monochromatic CT numbers of the contrast medium created by raw-data-based, dual-energy CT could be sufficient in clinical conditions.

INTRODUCTION

Dual-energy CT (DECT) is able to yield material-specific information using material decomposition algorithms that analyse the change in tissue attenuation between low- and high-energy scans.^{1,2} While mono-energy CT images can only provide CT number information, with DECT one can obtain various other data such as the effective atomic number (Z_{eff}), electron density and mono-energy CT number.^{3,4} DECT can also reduce beam-hardening artefacts and provide more quantitatively accurate attenuation measurements, which are considered to be the main benefits of virtual monochromatic DECT images. Thus, DECT has been applied in many clinical areas, including

the automatic characterization of stone compositions, bone removal and the diagnosis of gout.^{5–7} There are four types of DECT scanners that differ in the technique used to acquire high- and low-energy CT data sets: a dual-source, dual-energy scanner; a single-source, dual-energy scanner with fast kilovoltage switching; a single-source, dual-energy scanner with dual-detector layers; and a single-source CT that switches voltages between gantry rotations. In this study, we used a single-source CT that switches voltages between gantry rotations, as implemented in Toshiba Aquilion ONE™ (Toshiba Medical Systems Corporation, Ōtawara-shi, Japan). An advantage of this approach is the perfect alignment of the subsequent images, allowing

material decomposition to be performed in the raw projection space domain. Other approaches use image-based reconstruction for the material decomposition.

Mitchell et al reported on the accuracy of monochromatic CT numbers and Z_{eff} values that were calculated with high- and low-kV CT images using a GE Discovery CT750 DECT scanner (GE Healthcare, Princeton, NJ).⁸ Additionally, the use of iodine mapping with DECT has been shown to be advantageous for many clinical studies.⁹ An iodine distribution map can be used to detect perfusion defects in the case of liver embolism, pulmonary embolism and myocardial perfusion. Moreover, an iodine distribution map can further contribute to lesion detection and tissue characterization.¹⁰ Lee et al reported that quantifying iodine concentration has the potential to distinguish between different types of tumours.¹¹ To the best of our knowledge, a few studies have been conducted to examine the accuracy of iodine concentration using DECT.^{12,13} However, these studies in iodine mapping relied exclusively upon software-calculated, image-based reconstruction.

In our study, we used a Toshiba Aquilion ONE DECT scanner and evaluated the accuracy of the monochromatic CT numbers and Z_{eff} values created by raw-data-based decomposition of the DECT data for contrast material of varying iodine concentrations. We then compared these with the theoretical values.

METHODS AND MATERIALS

Phantom

An Electron Density Phantom Model 062M (Computerized Imaging Reference Systems, Inc., Norfolk, VA), as shown in Figure 1, was used. The phantom contained several tissue-equivalent inserts: lung (inhale), lung (exhale), adipose, breast, water, muscle, liver, trabecular bone (200 mg cc⁻¹ hydroxyapatite), dense bone (800 mg cc⁻¹ hydroxyapatite) and dense bone (1250 mg cc⁻¹ hydroxyapatite), whose atomic compositions and densities are well-known and provided by the manufacturer. Figure 1 shows the acrylic phantom, into which varying concentrations of iodinated contrast medium were inserted. Syringes were filled with solutions prepared by diluting iodine contrast medium (Omnipaque 300, GE Healthcare, Princeton, NJ) with water to predetermined concentrations of 0, 1, 2, 3, 4, 5, 6, 7, 8, 9, 10, 20, 40, 60, 90 and 130 mg iodine per ml. The syringes were positioned

around the centre. Figure 1 shows axial CT images of the iodine syringes.

Data acquisition

DECT studies were performed with a 320-detector CT scanner. CT scans were performed at tube voltages of 80 and 135 kV using the volume scanning method. Exposures of 800 and 200 mA were used to minimize noise. The time taken to switch the tube voltage between 80 and 135 kV was 0.4 s. The other scanning parameters were a rotation time of 1.0 s, slice thickness of 0.5 mm and field of view of 400 mm.

Z_{eff} values and monochromatic CT numbers reconstructed using raw-data-based DECT

Although the peak kilovoltage represents the upper limit of the energy spectrum, other energies exist between the upper and lower limits. The DECT scanner is able to generate monochromatic images from material-specific images by using a complex algorithm. The resultant monochromatic CT images, which are post-processed from a dual-energy data set, depict objects as if they were imaged with a monochromatic beam, and the X-ray energy is measured in kiloelectron volts (keV) instead of peak kilovoltage (kVp). Moreover, the electron densities and Z_{eff} values of each material were calculated using raw data decomposition from the DECT data. In the diagnostic energy range, the linear attenuation coefficient μ of a material can be described in photoelectric absorption and Compton scattering (excluding K-edge effects) as follows:

$$\mu(E) = a_p f_p(E) + a_c f_c(E) \quad (1)$$

where E is the photon energy, a_p and a_c are constants that depend on the material and f_p and f_c are constants that depend on the E values of photoelectric absorption and Compton scattering, respectively.^{14, 15} This formula can be simplified as follows:

$$\mu(E) = f(\rho_e, Z, E) + f(\rho_e, E) \quad (2)$$

where ρ_e is the electron density and Z is the effective atomic number. In dual-energy processing, a linear attenuation coefficient can be expressed by two basis materials as follows:

$$\mu(E) = \mu_1(E)c_1 + \mu_2(E)c_2 \quad (3)$$

where $\mu_1(E)$ and $\mu_2(E)$ are known functions of photon energy. The coefficients c_1 and c_2 are independent of energy and vary spatially, respectively. By solving equations (2) and (3), the electron density and Z_{eff} can be obtained. The corresponding monochromatic CT numbers in Hounsfield units (HU) are then computed using

$$CT\#(E) = \left[\frac{\mu_1(E)c_1 + \mu_2(E)c_2 - \mu(E)_{\text{water}}}{\mu(E)_{\text{water}}} \right] \times 1000 \quad (4)$$

where $\mu(E)_{\text{water}}$ is the linear attenuation coefficient of water at energy E . The total number of slices was 80, and the middle of these slices were analysed in all cases. The software package ImageJ (National Institutes of Health, Bethesda, MD) was used to

Figure 1. (a) Electron Density Phantom Model 062M, (b) Acrylic phantom, into which were inserted varying concentrations of iodinated contrast medium.

display the Z_{eff} image and monochromatic CT image. The mean value SDs of the Z_{eff} values and the mono-energy CT numbers were measured within a manually drawn region of interest using ImageJ. Circular regions of interest covering an area as large as possible within each of the materials that were used.

Theoretical Z_{eff} values and monochromatic CT numbers: The theoretical Z_{eff} values for the inserts in the CIRS 062M phantom were calculated using Mayneord's equation, as follows:¹⁶

$$Z_{\text{eff}} = 2.94 \sqrt{\sum_{i=1}^n a_i Z_i^{2.94}} \quad (5)$$

where a_i represents the fractional contribution of the i -th element to the total number of electrons in the mixture and Z_i is the atomic number. The theoretical monochromatic CT numbers of the inserts in the phantoms were computed from the mass attenuation coefficients determined using the NIST XCOM computer program (National Institute of Standards and Technology, Gaithersburg, MD), and from the mass densities supplied by the manufacturers. The chemical formulas of the mass fractions of the elements in the tissue-substitute inserts of the CIRS 062M phantom and in the iodinated contrast material inserts of the acrylic phantom were entered into XCOM, which computed the total mass attenuation coefficients of the inserts. The equation used to compute the monochromatic CT numbers of the inserts in both phantoms is as follows:

$$CT\#_i(E) = \left[\frac{(\frac{\mu}{\rho})(E)_i \times \rho_i - (\frac{\mu}{\rho})(E)_{\text{water}} \times \rho_{\text{water}}}{(\frac{\mu}{\rho})(E)_{\text{water}} \times \rho_{\text{water}}} \right] \times 1000 \quad (6)$$

where $(\mu/\rho)(E)_i$ is the mass attenuation coefficient of insert i at energy E , $(\mu/\rho)(E)_{\text{water}}$ is the mass attenuation coefficient of water at energy E , ρ_i is the mass density of insert i as supplied by the manufacturer and ρ_{water} is the density of water. Both measured and theoretical monochromatic CT numbers were obtained at monochromatic energies of 40–130 keV in increments of 10 keV.

RESULTS

Effective atomic number

The average measured Z_{eff} values, the theoretical Z_{eff} values and the deviation between them in the CIRS 062M phantom are shown in Figure 2. The average difference between the Z_{eff} values

Figure 2. (a) The average measured Z_{eff} values and the theoretical Z_{eff} values, (b) the deviation between them in the CIRS 062M phantom.

Figure 3. (a) The average measured Z_{eff} values and the theoretical Z_{eff} values, (b) the deviation between them in the acrylic phantom, into which were inserted varying concentrations of iodinated contrast medium.

of lung (inhale) inserts in the CIRS 062M phantom was 81.3% and, aside from these, the average difference between the Z_{eff} values was within 8.4%. Figure 3 shows the average measured Z_{eff} values, the theoretical Z_{eff} values and the deviation between them in the acrylic phantom, into which were inserted varying concentrations of iodinated contrast medium. The average difference between Z_{eff} values for contrast medium concentrations up to 90 mg ml⁻¹ was within 8.3%, and it exceeded 10% (11.2%) for concentrations greater than 130 mg ml⁻¹ for which the CT number was 3230 HU with 135 kV. The maximum SD of the contrast medium was 8.3% at 0 mg ml⁻¹.

Monochromatic CT number

Figure 4 shows the deviations between the measured and theoretical monochromatic CT numbers in the CIRS 062M phantom. The deviation between the measured and theoretical monochromatic CT numbers, except for bone inserts, in the CIRS 062M phantom showed good agreement within ± 80 HU. However, the bone inserts in the CIRS 062M phantom and the contrast medium phantom showed large deviations and the values were within ± 300 HU. For monochromatic energy, there were large deviations at less than 60 keV and the most significant difference occurred at 40 keV in all material inserts. Figure 5 shows the deviations between the measured and theoretical monochromatic CT numbers in the acrylic phantom, into which were inserted varying concentrations of iodinated contrast medium. The deviation between monochromatic CT numbers in the contrast medium was within 200 HU for concentrations less than 20 mg ml⁻¹. However, the average difference in monochromatic CT numbers in the contrast medium was above 300 HU at more than 20 mg ml⁻¹. At monochromatic energy, there were large deviations at less than 60 keV and the most significant difference occurred at 40 keV. Additionally, the smallest difference occurred at 70 keV.

DISCUSSION

This study revealed the accuracy of Z_{eff} values and monochromatic CT numbers reconstructed using raw-data-based

Figure 4. The deviations between the measured and theoretical monochromatic CT numbers in the CIRS 062M phantom.

DECT. The phantoms used different concentrations of iodinated contrast medium and human-tissue-equivalent materials. The measured Z_{eff} values of the lung (inhale) insert in the CIRS 062M phantom showed a large deviation from the theoretical Z_{eff} values. In a previous study, Mitchell et al reported that the Z_{eff} values of lung-simulating materials, calculated using CT images with DECT, could not be measured in either the Catphan (The Phantom Laboratory, Salem, NY) or Gammex 467 (Gammex Inc., Middleton, WI) phantoms.⁸ We measured the sinogram-based Z_{eff} value with DECT. As shown in Figure 2, there were large deviations and the SD was larger in the lung (inhale) insert than in the other inserts. The lung (inhale) insert includes little scattering material, and it mostly comprises non-scattering materials. Thus, it was difficult to measure accurately in the evaluation of the average value. Except for the lung (inhale) insert, the Z_{eff} values were accurate to within 8.4%. Mitchell et al reported that the Z_{eff} value calculated using the theoretical CT image with DECT was accurate to within 15% with Catphan phantom that was not used lung inserts.⁸ The raw-data-based Z_{eff} value was more accurate than that of the CT image-based Z_{eff} value. The average difference between the Z_{eff} values in the contrast medium exceeded 10% (11.2%) for concentrations over 130 mg ml⁻¹. The contrast medium included iodine, which is a high-atomic-number material. As the X-rays pass through the

body, low-energy X-ray photons are attenuated more easily, and the remaining high-energy photons are not attenuated as easily. Compared to low-atomic-number materials such as water, these high-atomic-number materials have an X-ray absorption edge at lower photon energies. Although the CT number at 135 kV for 130 mg ml⁻¹ contrast medium was over 3000 HU, there were no patients with such high CT numbers. Bae et al showed that the maximum CT number of abdominal aorta or liver that underwent an injection of contrast medium was within 400 HU and that the iodine concentration due to this was within 15 mg ml⁻¹ at 100–120 kV.¹⁷ Based on the results shown in Figure 3, the accuracy of the Z_{eff} values in the contrast medium showed good agreement, within 7.2%, at the concentration of 0–20 mg ml⁻¹. Although it was within the maximum SD in the contrast medium phantom, the SD in Z_{eff} images was large. It was the limitation of our study, and our future study will aim to create the new image filter or noise reduction technique such as iterative reconstruction in Z_{eff} image.

The bone inserts in the CIRS 062M phantom showed large deviations between the measured and theoretical monochromatic CT numbers as the density of the bone material increased, as shown in Figure 4. Moreover, as shown in Figure 5, the contrast medium phantom showed large deviations between the measured and theoretical monochromatic CT numbers as the concentration of iodine in the contrast medium increased. The deviation between the measured and theoretical monochromatic CT numbers in the contrast medium was above 300 HU at concentrations of more than 20 mg ml⁻¹, and this deviation was larger at lower monochromatic energy. The lower-energy image cannot eliminate the spectral beam-hardening artefacts. This effect was also larger as the concentration of the contrast medium increased. Therefore, the differences between the measured and theoretical Z_{eff} values and monochromatic CT numbers were larger for the high-concentration contrast media. To reduce this effect, we should use monochromatic CT images at energies over 70 keV. Montner et al found that the errors in the measured CT numbers depended on both the material being imaged and the monochromatic image.¹⁸ They reported that the inaccuracy of measured monochromatic CT numbers was lower at 80–100 keV. Our results show a similar trend, as shown in Figure 4. Montner et al showed the result of the errors in the measured CT numbers at 90 keV that were within 80 HU in tissue-equivalent phantom. In the current study, these errors were smaller within 25 HU at 90 keV. Thus, the raw-data-based monochromatic CT number were more accurate than that of the CT image-based monochromatic CT numbers. The minimum overall differences between the measured and theoretical monochromatic CT numbers of the tissue-equivalent phantom were at 70–130 keV. Additionally, the minimum overall differences between the measured and theoretical monochromatic CT numbers of the contrast medium phantom were at 70 keV. From the above results, the optimum monochromatic energy for human tissue and iodinated contrast medium is 70 keV. However, the monochromatic CT numbers of materials still depend on the amount of surrounding tissue—especially at low energies—thus demonstrating that the numbers are not truly monochromatic. Mono-energy also depends on patient body size. This is the limitation of our study, and further

Figure 5. The deviations between the measured and theoretical monochromatic CT numbers in the acrylic phantom.

research is needed to investigate more accurate monochromatic CT numbers and Z_{eff} values.

Although the iodine map in existing system could estimate the concentration of the contrast medium, it used only CT data.¹² The current study revealed the detailed material data of the Z_{eff} values and the monochromatic CT numbers of the human-tissue-equivalent and the iodinated contrast medium. There is a possibility that these data contribute to improvement of estimation accuracy of the contrast medium distribution using the Z_{eff} values and the monochromatic CT numbers in addition to conventional CT numbers. Future study should improve the image quality of the Z_{eff} values and seek to establish a method which distinguishes between the contrast

medium and other tissues using these various data for patients in clinical conditions.

CONCLUSION

The accuracy of raw-data-based Z_{eff} values was higher than that of image-based Z_{eff} values in the tissue-equivalent phantom. The accuracy of Z_{eff} values in the contrast medium showed good agreement within the maximum SD found within the iodine concentration range of clinical dynamic CT imaging. Moreover, the optimum monochromatic energy for human tissue and iodinated contrast medium was found to be 70 keV. The raw-data-based DECT data such as Z_{eff} values and monochromatic CT numbers could contribute the estimation of the contrast medium distribution and material decomposition in clinical data.

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