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Original paper

Image quality characteristics for virtual monoenergetic images using dual-layer spectral detector CT: Comparison with conventional tube-voltage images

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A R T I C L E I N F O

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A B S T R A C T

Purpose: To investigate the image quality characteristics for virtual monoenergetic images compared with conventional tube-voltage image with dual-layer spectral CT (DLCT).
Methods: Helical scans were performed using a first-generation DLCT scanner, two different sizes of acrylic cylindrical phantoms, and a Catphan phantom. Three different iodine concentrations were inserted into the phantom center. The single-tube voltage for obtaining virtual monoenergetic images was set to 120 or 140 kVp. Conventional 120- and 140-kVp images and virtual monoenergetic images (40–200-keV images) were reconstructed from slice thicknesses of 1.0 mm. The CT number and image noise were measured for each iodine concentration and water on the 120-kVp images and virtual monoenergetic images. The noise power spectrum (NPS) was also calculated.
Results: The iodine CT numbers for the iodinated enhancing materials were similar regardless of phantom size and acquisition method. Compared with the iodine CT numbers of the conventional 120-kVp images, those for the monoenergetic 40-, 50-, and 60-keV images increased by approximately 3.0-, 1.9-, and 1.3-fold, respectively. The image noise values for each virtual monoenergetic image were similar (for example, 24.6 HU at 40 keV and 23.3 HU at 200 keV obtained at 120 kVp and 30-cm phantom size). The NPS curves of the 70-keV and 120-kVp images for a 1.0-mm slice thickness over the entire frequency range were similar.
Conclusion: Virtual monoenergetic images represent stable image noise over the entire energy spectrum and improved the contrast-to-noise ratio than conventional tube voltage using the dual-layer spectral detector CT.

1. Introduction

Dual-energy computed tomography (CT) is performed using different scan techniques, including dual-spin, tube potential switching, and dual-source beam techniques, and two different X-ray spectrums are acquired [1–3]. These techniques have been used in valuable clinical applications, such as virtual monoenergetic imaging, iodine mapping, determination of effective atomic number, and measurement of electron density [4–9].

Recently, a raw data-based dual-energy CT called dual-layer spectral detector CT has become available [10–14] as a new device. Unlike tube voltage beam method, dual-layer spectral detector CT is the first commercially available detector-based CT that uses a single-tube voltage beam. With dual-layer spectral detector CT, low-energy photons are absorbed by the first detector layer, and high-energy photons are absorbed by the second. Therefore, dual-layer spectral detector CT not only provides valuable dual-energy information but is also routinely used in conventional tube-voltage images (e.g., 120 or 140 kVp) simultaneously.

So far, substantially increased image noise in lower monoenergetic images (e.g., 40–50 keV) has been a critical issue in clinical situations because increasing image noise degrades image quality, and

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deteriorated virtual monoenergetic images cannot provide a sustainable clinical image [15,16]. To decrease image noise of virtual monoenergetic images, dual-layer spectral detector CT introduces an “anti-correlated statistical reconstruction algorithm” in spectral reconstruction [17,18] which operates in the image and projection domains (post-decomposition), the latter being possible only in a projection-based decomposition technology (with a perfect alignment raw data sets) [19]. Then the method starts with converting a pair of high and low energy datasets obtained from dual-layer detector to Compton scatter and photoelectric data. It then performs spectral reconstruction, combining the following three points: the first point is maximizing a regularized log-likelihood. The second point is based on the fact that the noise of Compton and photoelectric data were negatively correlated. The noise can be effectively decreased by canceling out the correlated noise. The third point is using a synthesized data via a weighted sum of Compton and photoelectric data. This algorithm is effective in suppressing increased image noise by performing basis material decomposition on the spectral data. The spectral reconstruction for dual-layer CT made the amount of noise rather consistent regardless of the effective energies of monoenergetic keV images. Therefore, decreased image noise is expected in virtual monoenergetic images relative to the noise in other dual-energy mode images. In addition, a stable iodine CT number is also expected for virtual monoenergetic images regardless of the object size and acquisition method (120 or 140 kVp), and even conventional polychromatic image noise is decreased with increasing object size [4,20]. The improvement in the image quality with dual-layer spectral detector CT is expected to be routinely utilized for poor vessel enhancement, low iodine-contrast detectability, and so on. However, the image quality of dual-layer spectral detector CT virtual monoenergetic images is uncertain with respect to specific characteristics, such as image noise, iodine enhancement at different object sizes, and improvement of the contrast-to-noise ratio.

The purpose of this study was to investigate the image quality characteristics for comparison of virtual monoenergetic images with conventional tube-voltage images in a dual-layer spectral detector CT scanner.

2. Materials and methods

2.1. Phantom

For identifying the effect of iodine in different phantom sizes, two acrylic cylindrical phantoms with different size: small, 20 cm and 15 cm (diameter and height) and large, 30 cm and 15 cm were completely scanned. For our measurements, we made a concentration of iodinated enhancing material at 3.8 mg I/ml, 7.5 mg I/ml, and 15 mg I/ml (Iopamiron 300; Bayer Healthcare, Osaka). Each iodinated enhancing material was separately added along with distilled water to cylindrical silicon tubes of 1.0 cm (diameter) × 10 cm (length). The cylindrical silicon tube including the iodinated enhancing material or water was separately inserted into the small and large acrylic cylindrical phantoms in the same central position. A Catphan phantom (Phantom Laboratory, Cambridge, NY) with a CTP712 uniformity module was acquired to calculate the noise power spectrum (NPS) [21,22]. The diameter and height (length along the z-axis) of the phantom were 20.0 cm.

2.2. Helical scanning and image reconstruction

Helical scans were performed using a first-generation dual-layer spectral detector CT scanner (IQon Spectral CT; Philips Healthcare, Cleveland, OH). The single-tube voltage for obtaining virtual monoenergetic images was set to 120 or 140 kVp. Two acrylic cylindrical phantoms with different size and a Catphan phantom were completely scanned. The following scan parameters were used: detector configuration, 64 × 0.625 mm (detector collimation); gantry rotation time, 0.75 s; beam pitch, 0.8; and display field-of-view (FOV), 22.0 cm for 20-cm acrylic cylindrical and Catphan phantoms and 32 cm for a 30-cm acrylic cylindrical phantom. The tube current-time product was set to 110 mAs at 120 kVp, 80 mAs at 140 kVp for a 20-cm acrylic cylindrical phantoms and 800 mAs at 120 kVp, 550 mAs at 140 kVp for a 30-cm acrylic cylindrical phantom. Conventional 120- and 140-kVp images and virtual monoenergetic images (40–200-keV images) were reconstructed at slice thicknesses of 1.0 and at a slice interval of 1.0 mm with an abdomen standard kernel (C). In addition, the virtual monoenergetic images were also reconstructed using corresponding denoising levels 2 and 4(iDose^4 level 2 and 4), which decreased the image noise with increasing level.

2.3. Measurement of the CT number, image noise, and CNR

Using a slice thickness of 1.0 mm, the CT number was measured for each iodinated enhancing material in the acrylic cylindrical phantom on conventional 120- and 140-kVp, and virtual monoenergetic images (40–200 keV), including those images with denoising levels 2 and 4 (iDose^4 level 2 and 4). One of the authors (D.S) measured 20 consecutive images of the center region along the z-axis using a 4.0 mm-diameter region of interest (ROI) and calculated the mean value. Image noise was also measured at the center position using the inserted water object at a slice thickness of 1.0 mm, and the mean value was calculated from 20 consecutive images of the center region. Image noise was calculated as the root mean square value of the standard deviation of the CT numbers in the ROI of a 4.0-mm diameter. CNR values were calculated as follows:

$$\mathrm{CNR} = (\mathrm{ROI_m} - \mathrm{ROI_b})/\mathrm{SD_b}$$

where $\mathrm{ROI_m}$ was the mean CT number of the iodinated enhancing material, and $\mathrm{ROI_b}$ and $\mathrm{SD_b}$ are, respectively, the mean CT number of the water object and mean standard deviation of the CT numbers of the background [21,23]. In addition, the relative CNR was calculated by the CNR for each keV image divided by the CNR for 120-kVp images without denoising.

2.4. Noise power spectrum (NPS)

For the conventional 120-kVp images and each virtual monoenergetic image from 40, 70, and 100 keV, including denoising levels of 2 and 4 at slice thicknesses of 1.0 mm, 30 noise images along the z-axis were produced by subtracting the middle image. The 15 noise images from top slice levels and 15 from bottom slice levels (total 30 noise images) were chosen to avoid the noise correlation introduced by subtracting the middle image. There was 5-mm gap between the middle image and the first used image of top or bottom side. A central portion of the noise images with 256 × 256 pixels was chosen as the ROI, and the power spectrum of the ROI was obtained by the squared magnitude of the two-dimensional Fourier transform of the noisy ROI image divided by the physical area of the ROI [24,25]. The process was repeated for 30 noise images, and the mean of the spectrum over 30 noise realizations was calculated and defined as the NPS. In addition, the expected image noise for 120 kVp, 40 keV, and 70 keV was calculated by a square-root of the area-under-the-NPS-curve, which was then normalized against that of 70 keV.

3. Results

3.1. Iodine CT number

Fig. 1 shows the iodine CT numbers from 40- to 200-keV virtual monoenergetic images at different iodinated enhancing and phantom sizes acquired at 120 kVp (Fig. 1a) and 140 kVp (Fig. 1b). The iodine CT number at different iodinated enhancing was similar with respect to phantom size and acquisition method. At 15.0 mg I/ml and 120-kVp