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Research article

Noise characteristics of virtual monoenergetic images from a novel detector-based spectral CT scanner

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ABSTRACT

Aim: To evaluate the noise characteristics of virtual monoenergetic images (VMI) obtained from a recently introduced dual-layer detector-based spectral CT (SDCT), both in a phantom and patients.
Materials and methods: A cylindrical Catphan® 600 phantom (The Phantom Library, Salem NY, USA) was scanned using the SDCT. Image noise, signal-to-noise ratio (SNR), and contrast-to-noise ratio (CNR) were measured in VMI from 40 to 200 keV as well as conventional 120 kVp images. One hundred consecutive patients who had an abdominal CT on the SDCT were then recruited in the study. Noise, SNR and CNR were measured in the liver, pancreas, spleen, kidney, abdominal aorta, portal vein, muscle, bone, and fat, both in VMI (40–200 keV) and conventional 120 kVp images. Qualitative image analysis was performed by an independent reader for vascular enhancement and image quality on a 5 point scale (1-worst, 5-best).
Results: On phantom studies, noise was low at all energies of VMI. Noise was highest at 40 keV (5.3 ± 0.2 HU), gradually decreased up to 70 keV (3.6 ± 0.2 HU), after which it remained constant up to 200 keV (3.5 ± 0.2 HU). In the patient cohort, noise was low (< 25 HU) at all the energy levels of VMI for all the regions, with the exception of bone. For example, noise in the liver was highest at 40 keV (13.2 ± 4.6 HU), steadily decreased up to 70 keV (12.0 ± 4.4 HU) and then remained constantly low up to 200 keV (11.6 ± 4.3HU). For liver, pancreas, portal vein, aorta, muscle and fat, noise at all levels of VMI was lower than of conventional images ($p < 0.01$). For all organs, SNR, and CNR were highest at 40 keV (6.8–34.9; 18.3–44.9, respectively) after which they gradually decreased up to 120 keV (3.4–6.5; 9.5–13.0) and then remained constant to 200 keV (2.6–5.5; 8.5–12.5). Qualitative scores of VMI up to 70 keV were significantly higher than the conventional images ($p \le 0.01$), whereas for VMI ≥ 80 keV, they were lower than conventional images ($p < 0.001$).
Conclusion: VMI obtained from the novel SDCT scanner have low noise across the entire spectrum of energies. There are significant SNR and CNR improvements compared to conventional 120 kVp images.

1. Introduction

Dual energy CT (spectral CT/multi-energetic CT) utilizes two different energy spectra at acquisition to provide more detailed material characterization of various tissues, providing information beyond what is possible with conventional single energy CT. Once a predominantly research tool, dual energy CT is now routinely used in clinical practice [1–7]. Dual energy CT technologies operate either at the source or detector level, including dual source, rapid kVp switching, split beam, dual spin, multi-layer detectors, and photon-counting detectors. Several spectral images are generated from the dual energy technology by a process of two or three material decomposition, such as iodine map, virtual non contrast, effective atomic number and uric acid pair images [8]. Virtual monoenergetic images (VMI) are also generated, which mimic an x-ray beam composed of a single photon energy [9]. VMI at low energies are useful in enhancing vascular contrast due to higher photoelectric attenuation as the energies approach K-edge of iodine [10] and in improving lesion conspicuity [11,12]. VMI at higher energy levels have lower vascular contrast but are useful in reducing several artifacts such as beam hardening, calcium blooming and metallic artifacts [9,13].

VMI is generated by a weighted combination of photoelectric and Compton scatter basis images, and during the process of decomposition into these basis images, anti-correlated noise is introduced. Increased

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noise in low-energy VMI has been shown in both phantom studies [14–18] and in various patient cohorts [19–21], which limits its utility and diagnostic capabilities. In a dual source scanner, high noise was shown both at low and high energy VMI, with the optimal low-noise energy varying on the patient size (68, 71, 74, 77 keV for small, medium, large and extra-large phantoms respectively) [17]. Similar results have been shown in rapid kVp switching scanners, with low noise between 67 and 72 keV (least at 69 keV), and higher noise both at low and higher ends of the energy spectrum [18]. Recently noise has been reduced in dual source scanners, by using a second-generation monoenergetic plus algorithm [22,23].

VMI from 40 to 200 keV are also generated by projection space decomposition from the recently introduced detector-based spectral CT (SDCT), which has two layers of detectors, with the top layer absorbing the low energy photons and the bottom layer absorbing the high energy photons [24–26]. In this study, we sought to evaluate the noise characteristics of these VMI from SDCT, both in phantom and patient studies.

2. Materials and methods

This study was a Health Insurance Portability and Accountability Act–compliant study approved by our institutional review board. Informed consent was obtained from all the patients. Patients younger than 18 years and pregnant women were excluded from the study.

2.1. Phantom experiment

Phantom studies were performed using a cylindrical Catphan® 600 phantom (The Phantom Library, Salem NY, USA). The low contrast module of this phantom (CTP515) has several cylindrical cords of 40 mm length and various diameters with three contrast levels (Fig. 1a). The targets in the phantom as well as background material have equivalent effective atomic numbers, but variable density to change the effective attenuation coefficients. The phantom was scanned ten times on a SDCT prototype scanner (Philips Healthcare, Cleveland, OH, USA) for better statistical representation of the mean values. Scanning parameters were as follows: 120 kVp, 158 mAs, 0.33 s gantry rotation time and 64 × 0.625 mm collimation.

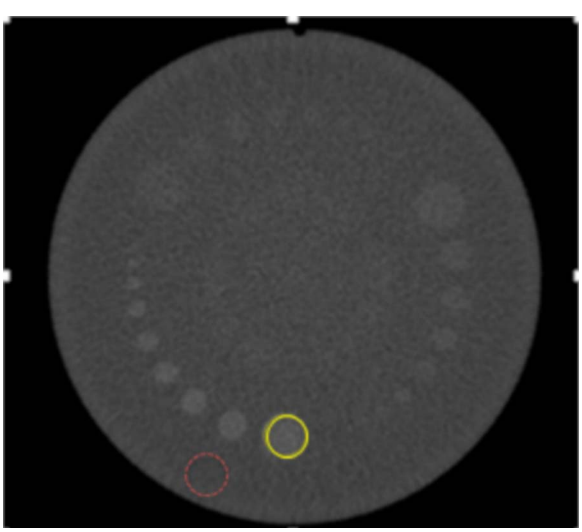

Fig. 1. Phantom study. CT image for evaluating the noise was obtained by scanning the CTP515 low contrast module of a Catphan® 600 phantom. An ROIc (yellow circle) was placed in the 1% cylindrical low-contrast target and ROIb of the same size (red circle) was placed in the background. Signal to noise ratio (SNR) was calculated as: HU_c/noise, and contrast to noise ratio (CNR) was calculated as $(HU_c - HU_b)$/noise.

2.1.1. Patient cohort

The study group compromised of 100 consecutive patients who had an abdominal CT scan in the SDCT scanner from October 2013 to October 2015. This included 53 routine CT abdomen with contrast, 31 CT angiography and 16 TAVI (transcatheter aortic valve implantation) studies. Examinations were performed for several indications, including assessment of abdominal pain, renal or liver mass evaluation, pre-renal and liver transplant assessment, vascular lesions and pre-aortic valve placement evaluation. The contrast dose and timing of contrast administration varied for the different examinations depending on the protocol, body mass index and renal function. Either Isovue 370 (Bracco Diagnostics Inc, Princeton, JU) or Ultravist 350 (Bayer Healthcare, Wayne, NJ) were used with contrast dose ranging from 40 to 150 ml. All patients were scanned using 120 kVp tube voltage with mAs adapted to the body size and automatic tube current modulation. Although based on BMI, some of these patients could have been scanned at 100 kVp in a conventional equivalent scanner, a tube voltage of at least 120 kVp is required for adequate spectral separation in this SDCT scanner. The mAs was reduced correspondingly in these patients to maintain dose neutrality with the conventional scanner. The detector configuration was 64 × 0.625 mm. The pitch ranged from 0.5 to 1.17 and gantry rotation time ranged from 0.3–0.75 s depending on the clinical indication. All the patients were scanned in the supine position. Some image sets also included scans of the chest, depending on the clinical indication.

2.1.2. Dual energy image processing and monoenergy creation

For both phantom and patient studies, conventional polyenergetic images at 120 kVp were generated by using combined data from both the spectral detector layers. These conventional polyenergetic images were reconstructed using iterative reconstruction algorithm (iDose[4] Level 3, Philips, Cleveland OH, USA) at 2 mm thickness with 1 mm overlap and a B (standard) filter. VMI were generated from spectral raw data using a dedicated workstation (Intellispace Portal, Philips Healthcare, The Netherlands). VMI were generated at 40, 50, 60, 70, 80, 100, 120, 140, 160, 180, and 200 keV energies, at 2 mm thickness with 1 mm overlap and B (standard) filter.

2.1.3. Image analysis

Image analysis was performed on a separate workstation (thin-client Spectral Diagnostic Suite, Philips Healthcare). For phantom images, a region of interest (ROI_c) was placed in the 15 mm, 1% cylindrical low-contrast target (Fig. 1) by an independent reader. Noise was calculated as the standard deviation of the pixel values in the ROI_c, and the mean Hounsfield Unit (HU) value (HU_c) was calculated as the mean pixel value. A background ROI (ROI_b) of the same size was also placed and the mean HU value (HU_b) was calculated as the mean pixel value within the ROI. Signal to noise ratio (SNR) was calculated as: HU_c/noise, and contrast to noise ratio (CNR) was calculated as $(HU_c - HU_b)$/noise. This analysis was repeated for all generated monoenergies.

For patient images, multiple ROIs were placed within the liver, pancreas, spleen, renal cortex, abdominal aorta, portal vein, paraspinal musculature, vertebral body, and subcutaneous fat by an independent reader with three years experience in CT image analysis. The size of each ROI was 1 cm^2, except in the smaller structures, in which case the largest possible ROI was placed. The signal was calculated as the mean HU within the ROI and noise was calculated as the standard deviation of the pixel values. SNR was calculated as: HU/Noise, and CNR was calculated as $(HU - HU_{fat})$/Noise for each tissue. This analysis was repeated for VMI at all energy levels. The effective diameter of the abdomen was measured at the location of ROI measurements and calculated as the square root of the product of the anteroposterior and transverse diameters. The patients were divided into three groups for sub analysis-small (less than 28 cm); medium (28–33 cm) and large (greater than 33 cm).

Qualitative image analysis was subsequently performed independently