Tin-filter Enhanced Dual-Energy-CT:

Image Quality and Accuracy of CT Numbers in Virtual Noncontrast Imaging

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Objectives: To measure and compare the objective image quality of true noncontrast (TNC) images with virtual noncontrast (VNC) images acquired by tin-filter–enhanced, dual-source, dual-energy computed tomography (DECT) of upper abdomen.

Materials and Methods: Sixty-three patients received unenhanced abdominal CT and enhanced abdominal DECT (100/140 kV with tin filter) in portal-venous phase. VNC images were calculated from the DECT datasets using commercially available software. The mean attenuation of relevant tissues and image quality were compared between the TNC and VNC images. Image quality was rated objectively by measuring image noise and the sharpness of object edges using custom-designed software. Measurements were compared using Student two-tailed *t*-test. Correlation coefficients for tissue attenuation measurements between TNC and VNC were calculated and the relative deviations were illustrated using Bland-Altman plots.

Results: Mean attenuation differences between TNC and VNC ($HU_{TNC} - HU_{VNC}$) image sets were as follows: right liver lobe −4.94 Hounsfield units (HU), left liver lobe −3.29 HU, vena cava −2.19 HU, spleen −7.46 HU, pancreas 1.29 HU, fat −11.14 HU, aorta 1.29 HU, bone marrow 36.83 HU (all $P < .05$); right kidney 0.46 HU, left kidney 0.56 HU, vena portae −0.48 HU and muscle −0.62 HU (nonsignificant). Good correlations between VNC and TNC series were observed for liver, vena portae, kidneys, pancreas, muscle and bone marrow (Pearson's correlation coefficient ≥0.75). Mean image noise was significantly higher in TNC images ($P < .0001$). Measurements of edge sharpness revealed no significant differences between VNC and TNC images ($P = .19$).

Conclusion: The Hounsfield units in VNC images closely resemble TNC images in the majority of the organs of the upper abdomen (kidneys, liver, pancreas). In spleen and fat, Hounsfield numbers in VNC images are tend to be higher than in TNC images. VNC images show a low image noise and satisfactory edge sharpness. Other criteria of image quality and the depiction of certain lesions need to be evaluated additionally.

Key Words: Dual energy; dual source; computed tomography; virtual noncontrast imaging.

The idea of dual-energy computed tomography (DECT) has existed since the invention of CT (1). However, only recently CT technology has advanced far enough to enable dose-neutral DECT acquisitions (2). Material differentiation in DECT is based on the dependence of photo absorption on energy and atomic number. If CT data that were acquired with different energies are available, a differentiation of the effective atomic numbers of the scanned materials and thus a differentiation of materials with different atomic numbers is feasible (3–6). Besides other possible technical approaches, dual-source CT with its two x-ray tubes can supply two simultaneous CT acquisitions with different energy levels (2,3). The energetic difference between the two spectra can be enhanced using a tin filter (7). The filter leads to a sharpening of the high-energy spectrum and thus a higher mean energy level of this spectrum and more effective dual energy discrimination. Thus, it is possible to use of 100 kV instead of 80 kV to create the lower energy spectrum that leads to an improved image quality of the low-energy images especially in the abdomen (7–9). Several clinical applications of DECT have recently been investigated (7,10–15). Because of the high atomic number of iodine, the amount of iodine contributing to every voxel can be quantified in DECT datasets using a three-material decomposition algorithm (16,17). Thus, the iodine content can be subtracted from the contrast enhanced images, resulting in virtual noncontrast (VNC) images.

Nonenhanced scans can be of interest for the measurement of contrast uptake of lesions, for imaging metabolic changes of liver tissue, and for the detection of calcifications, hematoma, and hemorrhagic cysts. However, because the unenhanced scan is not necessary in many patients, many radiology centers tend to abstain from acquiring the additional images for reasons of dose reduction. Unfortunately, once contrast is administered, the acquisition of noncontrast image is no longer possible in a reasonable time interval, which can hamper the interpretation of incidental findings.

The VNC technique has been assessed in several studies which exist in the literature (15,16,18–22). However, there are certain disagreements in these studies regarding the

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efficacy of the VNC technique in different organs. Furthermore, in the majority of these studies, a first-generation, dual-source CT scanner without tin filter has been used (15,17,18). Furthermore, all previous studies have focussed exclusively on the measurement of attenuation values and image noise. Because the VNC algorithm includes a noise filter, it is of interest to additionally evaluate edge sharpness in the VNC images.

The aim of this study thus was to assess the efficacy of iodine subtraction from DECT datasets that were acquired with a second-generation, dual-source CT system, comparing the measured attenuation values in large volumes of interest (VOI) of different tissues in VNC images to the gold standard, the true noncontrast (TNC) images. Furthermore, image quality of the VNC images was rated objectively in terms of image noise and edge sharpness of TNC and VNC images.

MATERIALS AND METHODS

Patient Population

The institutional review board approved this retrospective study; informed consent was waived. The hospital radiology information system was queried to identify all patients who had received an abdominal CT examination including unenhanced CT and contrast-enhanced DECT. Typical indications were lesions in the kidney, pancreas and liver, kidney stones, and general tumor staging examinations. Between October 2010 and January 2011, 63 consecutive patients (35 women, 28 men; mean age 63 ± 12 years; age range 33–91 years) meeting these criteria were found and retrospectively included into the study. Exclusion criteria were big renal or hepatic lesions that make measurements of regular parenchyma impossible and severe hemochromatosis that might influence the efficacy of the material decomposition algorithm. No eligible patients met the exclusion criteria.

CT Scanning Protocol

All examinations were performed in clinical routine using a second generation DSCT system (SOMATOM Definition Flash, Siemens Healthcare, Forchheim, Germany). The system consists of two x-ray tubes mounted on the same gantry at an angle of approximately 90° that simultaneously revolve around the patient. Patients were positioned in the isocenter of the table in a supine position. Only patients with a body habitus that allowed the complete inclusion of the relevant organs in the smaller field of view (FOV) of the second detector (32 cm) are supposed to receive DECT in our institution. At first, the TNC images of the abdomen were acquired in single energy mode using a tube voltage of 120 kVp. The tube current time product was manually adapted to patient weight by the technician (patient weight <60 kg: 100 mAs; patient weight 60–90 kg: 120 mAs; patient weight >90 kg: 140 mAs). A detector configuration of 128 × 0.6 mm and a pitch of 0.6 were used. All TNC scans were acquired in inspiratory breathhold and planned from the dome of the liver to the iliac crest. Subsequently, an automatic doublehead power injector was used for the intravenous injection of nonionic iodinated contrast material (Ultravist 370, Bayer Vital GmbH, Leverkusen, Germany) at a flow rate of about 2.2 mL/second, followed by a saline chaser at the same flow rate. Contrast material dose was adapted to patient weight (80 mL for patients weighting less than 60 kg, 100 mL for patients weighting between 60 and 90 kg, and 120 mL for patients weighting over 90 kg). The portal-venous phase CT data acquisition was started 80 seconds after initiation of the contrast media administration. Basic DE scanning parameters were 100 and 140 kVp with additional tin filtration filter. Collimation and pitch were 128 × 0.6 mm and 0.7, respectively. Online dose modulation (CAREDose4D, Siemens Healthcare) was applied with quality reference mAs value of 151 mAs for 140 kVp and 196 mAs for 100 kVp.

Image Reconstruction and Postprocessing

After the examination, axial images were reconstructed using a section thickness of 1.5 mm and an increment 1.0 mm with a B30f reconstruction kernel for the single-energy scans and a D30f kernel for the DE scans. The D30f kernel is optimized for DE postprocessing and is equivalent to the B30f kernel, which is used in single-energy CT imaging regarding the image noise. The reconstructed FOV was individually adapted to patient size and identical in TNC and VNC reconstructions. Pixel Matrix was 512 × 512. VNC images were created using a dedicated commercial software (Liver VNC, syngo multimodality workplace, version VE40A, 2010, Siemens Healthcare).

Measurement of Hounsfield Numbers for TNC and VNC

After generation of VNC images, TNC and VNC images were registered with dedicated hybrid imaging software (syngo MMWP Oncology Engine with TrueD, VE32B, 2008, Siemens Healthcare) to compensate for movement artifacts in between the scans because of differences in breathhold. Manual adjustments of the registration that include rotation and panning were performed to achieve an optimal registration. In each study, VOIs were placed in 12 different anatomical locations (Table 1). Because the registration, attenuation values could be recorded from exactly corresponding VOIs in TNC and VNC images. In every organ, three VOIs were assessed and averaged. At all anatomical regions, a maximum possible size was obtained for the VOIs (mean size 7.24 cm^3, range 0.27–27.3 cm^3).

Objective Measurements of Image Quality

Examinations were rated acceptable for diagnostic purpose; if not relevant, incomplete inclusion of organs of the upper abdomen was observed.