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Research article

Impact of noise-optimized virtual monoenergetic dual-energy computed tomography on image quality in patients with renal cell carcinoma

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ABSTRACT

Objective: The aim of this study was to evaluate the impact of a noise-optimized virtual monoenergetic imaging (VMI +) reconstruction technique on image quality and lesion delineation in patients with renal cell carcinoma (RCC) undergoing abdominal dual-energy computed tomography (DECT).

Materials and methods: Fifty-two patients (33 men; 61.5 ± 13.6 years) with RCC underwent contrast-enhanced DECT during the corticomedullary and nephrogenic phase of renal enhancement. DECT datasets were reconstructed with standard linearly-blended (M_0.6), as well as traditional virtual monoenergetic (VMI) and VMI + algorithms in 10-keV increments from 40 to 100 keV. Contrast-to-noise (CNR) and tumor-to-cortex ratios for corticomedullary- and nephrogenic-phase images were objectively measured by a radiologist with 3 years of experience. Subjective image quality and RCC delineation were evaluated by three independent radiologists.

Results: Greatest CNR values were found for 40-keV VMI+ series in both corticomedullary- (8.9 ± 4.9) and nephrogenic-phase (7.1 ± 4.6) images and were significantly higher compared to all other reconstructions ($P < 0.001$). Furthermore, tumor-to-cortex ratios were highest for 40-keV nephrogenic-phase VMI+ (2.1 ± 3.5; $P \leq 0.016$), followed by 50-keV and 60-keV VMI+ (2.0 ± 3.2 and 1.8 ± 2.8, respectively). Qualitative image quality scored highest for 50-keV VMI+ series in corticomedullary-phase reconstructions and 60-keV in nephrogenic-phase reconstructions ($P \leq 0.031$). Highest scores for lesion delineation were assigned for 40-keV VMI+ reconstructions ($P \leq 0.074$).

Conclusion: Low-keV VMI+ reconstructions lead to improved image quality and lesion delineation of corticomedullary- and nephrogenic-phase DECT datasets in patients with RCC.

1. Introduction

Renal cell carcinoma (RCC) is the 9th most common cancer in men and the 14th most prevalent malignant tumor in women [1]. Although mortality due to RCC has been slightly decreasing over time, this disease still ranks as the 16th most common cause of death from cancer worldwide [1,2]. Clear cell and papillary cell carcinoma are the two most frequently encountered subtypes of RCC, and they account for approximately 70% and 20% of all cases, respectively [3,4]. The lack of early symptoms results in a high proportion of patients already having metastases and, moreover, nearly half of RCC cases are detected incidentally on radiographic examinations [5]. Moreover, CT has become increasingly important for staging and follow-up of patients with RCC.

Several prior studies demonstrated various advantages of dual-energy CT (DECT) for imaging and assessment of patients with renal lesions. It has been shown that virtual non-contrast images can replace a true unenhanced acquisition without deteriorating diagnostic accuracy and thus reduce radiation dose [6,7]. DECT-based iodine maps have been shown to allow for renal lesion characterization beyond the capabilities of traditional single-energy CT [8–10]. Furthermore, virtual monoenergetic imaging (VMI) may overcome renal cyst pseudoenhancement and thus allow for a more accurate differentiation between renal cysts and solid masses [11,12]. A noise-optimized virtual monoenergetic reconstruction algorithm (VMI +) has also recently been introduced to optimize image quality at low keV levels providing a distinctly increased iodine signal at simultaneously moderate noise

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levels unlike its predecessor technique [13,14]. In prior studies, this reconstruction method has shown favorable results for vascular and oncologic DECT [13,15–18]. To date, however, the application of VMI algorithms has not been investigated for patients with RCC lesions.

Therefore, the purpose of this study was to evaluate the impact of the noise-optimized VMI+ reconstruction technique on image quality and lesion delineation in patients with RCC undergoing abdominal DECT.

2. Materials and methods

2.1. Patient population

This retrospective single-center study was approved by the institutional review board of our university hospital with a waiver for written informed consent. Patient selection was performed by a radiologist with 5 years of experience in abdominal CT imaging who was not involved in the subsequent data analysis. Our picture archiving and communication system (PACS) databases were screened using keywords related to RCC to identify patients with histologically proven RCC and corresponding findings in DECT examinations between September 2014 and October 2016. The query yielded data of 69 patients who had undergone DECT on the same CT platform. Patients with deviations from the standard contrast media injection protocol (n = 11); severe motion artifacts (n = 2); and with datasets not available for retrospective image reconstructions (n = 4) were excluded.

The final study group included 52 patients (mean age, 61.5 ± 13.6 years; range, 31–87 years), consisting of 33 men (mean age, 60.2 ± 13.5 years; range, 31–82 years) and 19 women (mean age, 63.7 ± 13.6 years; range, 39–87 years).

2.2. DECT image acquisition

All dual-energy multidetector CT examinations were performed using a third-generation dual-source 192-row multidetector DECT scanner (SOMATOM Force, Siemens Healthcare, Forchheim, Germany). Image acquisition was performed in craniocaudal direction and inspiratory breath-hold. A triple-phase contrast-enhanced CT protocol was applied, consisting of image acquisition during a non-contrast, a corticomedullary, and a nephrogenic phase of enhancement.

The corticomedullary-phase scan was automatically started by using a dedicated 120-kV bolus tracking scan software (CareBolus, Siemens), 15 s after a threshold of 120 Hounsfield units (HU) was measured within a region of interest (ROI) in the descending aorta at the level of the celiac artery. The nephrogenic phase started with a delay of 90 s after the start of the contrast material injection [3]. A nonionic contrast agent (Imeron 400 mg iodine/ml, Bracco, Milan, Italy) was administered at a dose of 1.2 ml per kilogram body weight with a flow rate of 3 ml/s through an intravenous catheter inserted into an antecubital vein. Settings for the DECT mode were as follows: tube A 100kV, reference current-time product 95 mAs per rotation and tube B Sn150 kV with tin filter, 59 mAs per rotation [17]. Furthermore, rotation time was 0.5 s, the pitch was set to 0.7, and the collimation to 2 × 192 × 0.6 mm. All images were obtained using an automated exposure control system (CARE Dose 4D, Siemens). The volume CT dose index ($CTDI_{vol}$) and the dose length product (DLP) of each patient were recorded for an estimation of the DECT radiation dose.

2.3. DECT image reconstruction

DECT raw data were post-processed on a 3D multi-modality workstation (syngo.via, version VA30A, Siemens) using a soft tissue convolution kernel (Qr40, Siemens) and iterative reconstruction technique (ADMIRE, Siemens; strength level, 3). The default scanner software automatically generates linearly-blended images by merging 60% of the low-kV data and 40% of the high-kV data according to the vendor's recommendation. These images resemble standard 120-kV images of the abdomen [15,17,19]. The traditional VMI and noise-optimized VMI + series were reconstructed at 40, 50, 60, 70, 80, 90, and 100 keV levels. Images at higher energy levels beyond 100 keV were not calculated, as the iodine attenuation can be expected to be too faint [20]. All images were reconstructed as axial and coronal slices, with a thickness of 1.5 mm and increment of 1.0 mm, respectively.

2.4. Quantitative image analysis

For the evaluation of quantitative image quality, image series were reviewed by a radiologist with 3 years of experience in CT (S.S.M), who did not participate in the subsequent qualitative image interpretation. Signal attenuation of the renal cortex and RCC lesion in mean HU and image noise, defined as the standard deviation (SD) of fat, were measured in all patients. Measurements were performed by placing a circular region-of-interest (ROI) in the renal cortex (100 mm^2) and the RCC lesion (100 mm^2). Focal areas of tumor necrosis or heterogeneity were avoided. Additional measurements were performed within the psoas muscle (250 mm^2) and subcutaneous fat at the lower back (150 mm^2) to assess image contrast and background noise. Measurements were performed twice and averaged to ensure data consistency and reduce measuring inaccuracies.

In order to calculate signal-to-noise (SNR) and contrast-to-noise ratio (CNR) values in corticomedullary- and nephrogenic-phase images (later on referred to as SNR_C, CNR_C, SNR_N, and CNR_N), the following formulas were used according to previous studies [14,17,21]:

$$SNR = HU\ (tumor)\ /\ SD\ (fat)$$

$$CNR = (HU\ (tumor) - HU\ (muscle))\ /\ SD\ (fat)$$

Furthermore, the tumor-to-cortex ratio was calculated for corticomedullary- and nephrogenic-phase images using the following formula in conformity with previous studies [22,23]:

$$Tumor\ to\ cortex\ ratio = (HU\ (cortex) - HU\ (tumor))\ /\ SD\ (cortex)$$

2.5. Qualitative image analysis

Three radiologists with 3–6 years of experience in oncologic CT imaging (D.L., L.L., J.L.W.) independently assessed all DECT images and were blinded to the used reconstruction technique. Furthermore, readers were allowed to freely modify the preset window settings (width, 400 HU; level, 100 HU) as low-keV VMI+ reconstructions may require different width and level settings to improve the visualization and contrast conditions in abdominal DECT [24]. The order of the different image series was randomized, and only a single image series of each patient was evaluated during each reading session. Also, a time interval of at least two weeks was kept between each read-out to prevent potential recall bias. Observers were aware that all patients were diagnosed with histologically-proven RCC. Image series were rated using 5-point Likert scales under the following aspects [16]: Overall image quality (ranging from 1 = non-diagnostic image quality to 5 = excellent image quality), lesion delineation (ranging from 1 = poor visual delineation to 5 = excellent delineation of contours), and image noise (ranging from 1 = extensive image noise to 5 = absence of noise).

2.6. Statistical evaluation

Dedicated statistical software for biomedical research (MedCalc Statistical Software Version 17.2, MedCalc Software bvba, Ostende, Belgium) was used for all computations. Variables are expressed as means ± standard deviation. The Kolmogorov-Smirnov test was applied to test for normality of data distribution. Data showing a normal