European Journal of Radiology 104 (2018) 108–114

Contents lists available at ScienceDirect

European Journal of Radiology

journal homepage: www.elsevier.com/locate/ejrad

Research article

Dual-layer spectral computed tomography: Virtual non-contrast in comparison to true non-contrast images

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ARTICLE INFO

Keywords:
CT
Dual-Layer CT
Spectral CT
Material decomposition
Virtual-non-contrast
Iodine quantification

ABSTRACT

Purpose: To evaluate virtual-non-contrast (VNC) images obtained from clinical triphasic scans with a dual-layer spectral computed tomography system regarding accuracy of iodine subtraction.

Material and methods: From September to December 2016, 62 consecutive patients who underwent a clinical routine triphasic CT examination were included into this retrospective study. VNC images based on the arterial and portal venous phase were generated. For every patient and every contrast phase, a region-of-interest (ROI) was defined in aorta, liver, renal cortex, spongious bone, fat, muscle and fluid (i.e. gallbladder, urinary bladder), resulting in 2170 ROIs. VNC images were compared to true-non-contrast (TNC) images regarding difference in attenuation. Consistency between VNC images obtained from the arterial and portal venous phase as well as the influence of the initial attenuation on respective VNC images were evaluated.

Results: Comparison of HU in VNC and TNC images showed a high accuracy of iodine elimination. Mean difference between TNC and VNC images was only 0.5 ± 8.5 HU and > 90% of all comparisons showed a difference of less than 15 HU. For all tissues but spongious bone, mean absolute difference between TNC and VNC images was below 10 HU. VNC images derived from the arterial and the portal venous phase showed excellent correlation. The quality of iodine removal in VNC images was not influenced by the original contrast enhancement. However, VNC images cannot be used for evaluation of iodine removal in bone as bone and iodine can hardly be differentiated via spectral CT.

Conclusion: VNC imaging in DL-CT is a promising tool for daily clinical routine. As non-enhanced CT images are essential in multiple clinical situations, the permanent availability of VNC images with dual-layer spectral CT will result in a substantial reduction of radiation exposure and an increased diagnostic value of monophasic contrast-enhanced CT scans.

1. Introduction

Computed tomography (CT) is an essential imaging modality in multiple clinical situations [1]. Most examinations are performed with intravenously applied contrast media to improve detectability of pathologies. Usually a portal venous phase is adequate. In some situations (e.g. detection of bleeding and differentiation of certain tumors) additional contrast phases are needed [2], leading to increased radiation exposure [3].

Lately, dual-energy CT (DE-CT) has been introduced into clinical routine and was investigated intensively [4]. In contrast to conventional CT systems, these systems use two x-ray tubes with different peak kilovoltages (kVp) (so called dual-source CT) or one tube with kVp switching technology to obtain spectral data [5,6]. Predominantly, dual-source CT (DS-CT) is used in clinical routine. With these systems, acquisition of spectral data by modification of the tube currents is possible. However, spectral data are not automatically acquired in every scan. E.g. an examination can be performed using a current of 100 kVp for both tubes (100 kVp/100 kVp tube setting) without acquisition of spectral information – or it can be performed using a current of 140 kVp for one tube and a current of 90 kVp for the other tube (140/90 kVp tube setting) for acquisition of spectral data. Multiple different tube settings are possible to obtain spectral data, e.g. 140/90 kVp or 150/100 kVp. With DE-CT technology, spectral imaging and in consequence material decomposition/quantification is possible [7–9]. Multiple applications for DE-CT exist [8–15]. Here

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https://doi.org/10.1016/j.ejrad.2018.05.007
Received 4 October 2017; Received in revised form 9 March 2018; Accepted 5 May 2018

discrimination of iodized blood/calcified plaques in contrast-enhanced CT and quantification of iodine seem particular important [16]. By quantification of iodine, iodine can theoretically be subtracted completely, resulting in virtual-non-contrast (VNC) images [17]. Compared to true-non-contrast (TNC) images which are obtained in a scan before application of contrast media, in VNC images iodine is merely subtracted mathematically. For DS-CT, VNC images showed good results regarding the reliability of iodine subtraction [17].

Recently, a novel type spectral CT system based on a dual-layer detector was introduced into clinical routine [18]. In this dual-layer CT (DL-CT), the upper layer of the detector absorbs low-energy photons and is permeable for high-energy photons which are absorbed by the lower layer [19]. Thus, low- and high-energy images are generated and by weighted summation, full spectrum images with spectral information are calculated. As described above, examination settings must be defined to obtain spectral data before starting the CT scan when using DS-CT system. DL-CT overcomes this drawback and automatically acquires spectral data in each scan without modification of scanning parameters. Thereby a full retrospective spectral evaluation of CT data is possible whereas with previous DE-CT systems, retroactive generation of spectral data was not possible. This is helpful when a non-enhanced contrast phase is needed for diagnostics but was not obtained initially. Additionally, every single energy x-ray tube could generate spectral data if the dual layer detector technology was installed.

As many CT scans are performed in clinical routine, radiation reduction is essential [20] because it is a serious concern due to its potential to cause malignant tumors and gonadal damage [21,22]. Approaches for dose reduction have found their way into clinical routine, for example shielding techniques, reduction of anatomical scan coverage, z-axis modulation and beam pre-filtration [23,24]. During the last years, especially iterative reconstruction techniques led to substantially reduced radiation doses [25,26]. With current hardware, further reduction of radiation dose by improved reconstruction algorithms is extremely difficult as minimum radiation is needed to generate appropriate image quality. In contrast, radiation exposure can efficiently be further reduced by making the non-enhanced phase dispensable which could be possible via VNC imaging.

The present study evaluated VNC images obtained from clinical triphasic scans with a novel dual-layer spectral CT regarding accuracy of iodine subtraction. For this purpose, VNC images assessed from an arterial and a portal venous phase were compared to TNC images.

2. Material and methods

2.1. Patient population

Institutional review board approval was obtained for this retrospective study including all protocols (Ethics Committee, Technische Universität München, München, Germany). Informed consent was waived by the institutional review board as no additional data besides clinical obtained images were used. All examinations were performed exclusively for diagnostic use to full extend and were performed only with clinical standard protocols. All patient data were completely anonymized at the beginning of the study.

From September to December 2016, 62 consecutive patients, 28–89 years of age (mean age 69.7 years), 47 males (28–89 years, mean 70 years) and 15 females (32–86 years, mean 69 years), were retrospectively enrolled in this study. No preselection regarding patient weight, age, sex or other characteristics was performed. Patients who received intravenous contrast medium shortly before the mentioned triphasic CT scan were not included as due to remaining contrast medium no real non-enhanced images could have been obtained. Main indications for examinations were acute bleeding, evaluation of endovascular aortic repair (EVAR) or staging of selected malignant tumors, e.g. pancreatic carcinoma. Mean dose-length-product (DLP) for all examinations was 1386.2 ± 1086.8 mGy*cm. By multiplication with the conversion factor of 0.015 for the trunk, a mean effective dose (ED) of 20.7 ± 16.3 mSv resulted [27,28].

2.2. Image acquisition

All patients underwent a triphasic CT scan with a dual-layer spectral CT (IQon; Philips Healthcare, Cleveland, OH, USA) that included a non-enhanced scan, an arterial phase and a portal venous phase. All scans were performed with a tube voltage of 120 kVp. Patients were placed in supine position on the scanner couch. After an anteroposterior scout, a non-enhanced contrast phase was obtained. An adjustment of the tube current based on the scout view (z-axis modulation) is included by default. Subsequently, a weight adapted amount (1.2 ml/kg with a maximum of 120 ml) of contrast agent (Ultravist 370 MCT, Bayer Vital GmbH, Leverkusen, Germany) followed by a 50 ml saline chaser was injected intravenously with a rate of 3.5 ml/s using a dual syringe injection system (Stellant, MEDRAD, Indianola, Pennsylvania). A bolus tracker was placed within a region-of-interest (ROI) in the descending aorta to ensure optimal contrast enhancement (threshold for scan start: 150 HU). After reaching the threshold, the scan for the arterial contrast phase was started automatically. The venous contrast phase was started 75 s after injection of the contrast medium. The scan was performed craniocaudally with a pitch of 0.9 and a 64 × 0.625 mm detector configuration. All data sets were reconstructed in axial view with slice thickness of 3 mm and a 512 image matrix.

VNC images of the arterial and venous phase were calculated using the vendor specific spectral workstation (IntelliSpace Portal (v. 8.0.2), Philips Healthcare, USA). Apart from VNC-calculation, no other image characteristics were changed. VNC images were reconstructed in 3 mm axial slices and were therefore identical to the arterial and portal venous images regarding image positioning and orientation.

2.3. Image analysis

ROIs were defined in the following areas: aorta, liver, kidney (renal cortex), fat (subcutaneous or mesenterial), paravertebral muscle, spongious bone (lumbar vertebra), fluid without uptake of contrast medium (i.e. gallbladder, urinary bladder, pleural effusion). Every ROI had an area of more than 1 cm^2. HU were measured in the non-enhanced phase, the arterial and the portal venous phase as well as in the VNC images of the arterial and the portal venous phase, respectively. Images were lined up next to each other and matched in z-direction. The ROI for every tissue was defined in the non-enhanced images and then copied into the other series. As the orientation in the non-enhanced, arterial und portal venous phase could vary due to breathing and movement of the patient, position and/or slice number of some ROIs had to be adjusted for exact anatomical analogy of the ROIs in all reconstructions by a radiologist with 3 years of experience in CT diagnostics (AS).

In total, 2170 ROIs were defined (7 tissues, 5 reconstructions, 62 patients). As every tissue was examined in every patient, the number of measured ROIs (n-values) is not shown in every table or figure. For every ROI, the mean HU and the corresponding standard deviation (SD) were measured. Fig. 1 shows an example of the different regions and sizes of the described ROIs.

2.4. Influence of VNC on the noise level

To determine if VNC reconstruction influences the noise level, the SD of every ROI was measured. By calculation of the mean of all ROIs in one reconstruction of each tissue, the noise in this reconstruction/tissue was quantified.

2.5. Statistical analysis

Statistical analysis was performed by dedicated software packages