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Contrast agent concentration optimization in CTA using low tube voltage and dual-energy CT in multiple vendors: a phantom study

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Abstract

We investigated the feasibility and extent to which iodine concentration can be reduced in computed tomography angiography imaging of the aorta and coronary arteries using low tube voltage and virtual monochromatic imaging of 3 major dual-energy CT (DECT) vendors. A circulation phantom was imaged with dual source CT (DSCT), gemstone spectral imaging (GSI) and dual-layer spectral detector CT (SDCT). For each scanner, a reference scan was acquired at 120 kVp using routine iodine concentration (300 mg I/ml). Subsequently, scans were acquired at lowest possible tube potential (70, 80, 80 kVp, respectively), and DECT-mode (80/150Sn, 80/140 and 120 kVp, respectively) in arterial phase after administration of iodine (300, 240, 180, 120, 60, 30 mg I/ml). Objective image quality was evaluated using attenuation, CNR and dose corrected CNR (DCCNR) measured in the aorta and left main coronary artery. Average DCCNR at reference was 227.0, 39.7 and 60.2 for DSCT, GSI and SDCT. Maximum iodine concentration reduction without loss of DCCNR was feasible down to 180 mg I/ml (40% reduced) for DSCT (DCCNR 467.1) and GSI (DCCNR 46.1) using conventional CT low kVp, and 120 mg I/ml (60% reduced) for SDCT (DCCNR 171.5) using DECT mode. Low kVp scanning and DECT allows for 40–60% iodine reduction without loss in image quality compared to reference. Optimal scan protocol and to which extent varies per vendor. Further patient studies are needed to extend and translate our findings to clinical practice.

Keywords Iodinated contrast media · Dual-energy CT · Thoracic aorta · Low kVp · Image quality

Abbreviations

AA	Ascending aorta
CNR	Contrast to noise ratio
CTA	Computed tomography angiography
$CTDI_{vol}$	Volumetric CT dose index
DA	Descending aorta
DCCNR	Dose corrected contrast to noise ratio
DECT	Dual-energy computed tomography
DSCT	Dual source computed tomography
FBP	Filtered back projection
GSI	Gemstone spectral imaging
IR	Iterative reconstruction
LM	Left main coronary artery
ROI	Region of interest
SD	Standard deviation
SDCT	Dual-layer spectral detector computed tomography
VMI	Virtual monochromatic imaging

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Introduction

Computed tomography angiography (CTA) is an important diagnostic tool for the detection of cardiovascular diseases [1]. However, while a powerful test, the use of CTA can be limited in patients with impaired renal function [2, 3]. Reducing iodine concentration decreases the risk of contrast-induced nephropathy [3, 4] and can therefore be beneficial in terms of safety and costs [5].

One of the methods to reduce the amount of iodine is scanning with low tube voltage. Several studies have described iodine dose reductions of up to 56% using

70–80 kVp acquisition while maintaining image quality compared to routine CTA [6–9]. At lower X-ray energy levels, attenuation of iodine contrast increases as the energy level approaches the K-edge of iodine (33.2 keV). Thereby allowing for iodine concentration reduction without decrease in CT-values [10].

An alternative method is the use of dual-energy CT (DECT) scanning. DECT discriminates X-ray photons at high and low energy levels and thereby allows for reconstruction of virtual monochromatic imaging (VMI) at discrete energy levels down to 40 keV. Several authors have reported iodine dose reductions of up to 70% with DECT for aortic and coronary CTA while maintaining objective image quality [11–15]. DECT became commercially available a decade ago. Since then multiple approaches have been introduced: (1) dual source CT (DSCT) with two X-ray tubes and two corresponding detectors, with each tube operating at a different tube voltage (low or high); (2) gemstone spectral imaging (GSI) with a single X-ray tube rapidly switching within one rotation between low and high voltage; (3) dual-layer spectral detector CT (SDCT) with a single X-ray tube and a dual-layer detector that differentiates one photon spectrum in a low and high energy fraction. Although these techniques all attempt to achieve the same, from a physics perspective, image data are acquired in a very different way and at different energy levels. In addition, image reconstruction methods to create VMI differ per vendor [16].

The aim of our study was to evaluate on a per vendor level the extent to which iodine concentration could be reduced by using low tube voltage and low VMI without loss of objective image quality compared to CT at routine tube voltage.
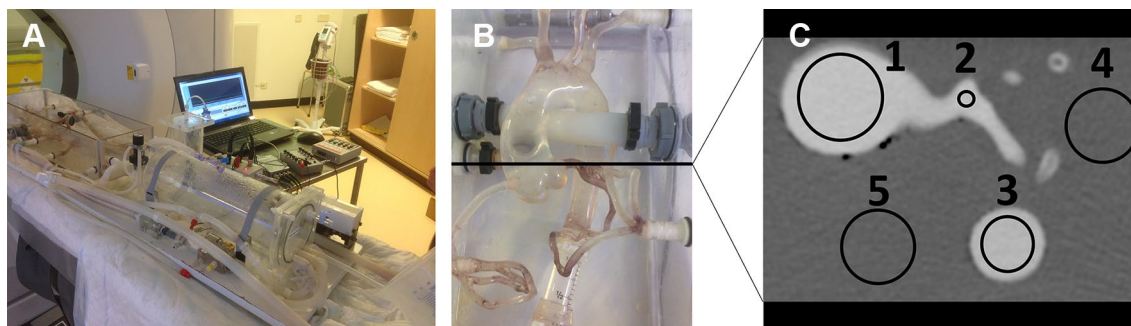
Materials and methods

Phantom setup

A dynamic circulation phantom with a low-pressure venous compartment and a high-pressure arterial compartment simulating physiological circulation parameters was used (Fig. 1a) [17, 18]. The phantom consisted of a water-filled (37 °C) acrylic container to mimic CT attenuation characteristics of the mediastinum. The container encased multiple connecting tubes, which accurately mimic the aorta and coronary arteries (Fig. 1b). A pressure relief valve simulates arterial and venous pressure. To mimic pulsation, a Harvard medical heart pump (BS4; Harvard Apparatus, Holliston, MA) was used. Physiological circulation was simulated by using the following settings: heart rate 60 beats per minute, stroke volume 60 mL per beat, output phase ratio 40% systole/60% diastole and blood pressure around 120/80 mmHg.

Injection protocol and image acquisition

Iodinated contrast medium of 300 mg I/ml (Iopromide, Ultravist 300, Bayer Healthcare, Berlin, Germany) was used and diluted with water to obtain the following concentrations (ratio contrast medium:water): 300(1:0), 240(4:1), 180(3:2), 120(2:3), 60(1:4) and 30(1:9) mg I/ml. A fixed injection protocol was used (volume 40 ml, flow rate 6 ml/s, injection time 6.7 s), resulting in an iodine delivery rate (IDR; concentration × flow rate) of 1.80, 1.44, 1.08, 0.72, 0.36 and 0.18 g I/s, respectively. All concentrations were preheated to 37 °C and injected into the phantom using a standard CT power injector (Stellant, MEDRAD, Pittsburgh, PA, USA) for both DSCT and SDCT, and a dual head power injector (Nemoto-Kyorindo, Tokyo, Japan) for GSI. After every scan,

Fig. 1 Dynamic circulation phantom setup. **a** Physiological circulation parameters were simulated using a low-pressure venous compartment and a high-pressure arterial compartment. **b** The phantom consists of a water-filled (37 °C) acrylic container that encased multiple connecting tubes, which accurately mimic the aorta and coronary arteries. **c** Intravascular enhancement and noise were measured by using a manually placed circular region of interest in the AA (*1*), LM (*2*), DA (*3*) and 2 homogeneous areas (*4* and *5*). *AA* ascending aorta, *DA* descending aorta, *LM* left main coronary artery

the phantom was flushed with water three times to remove residual contrast media.

Acquisitions were made using 3 DECT scanners; DSCT, Definition Force (Siemens Healthcare, Forchheim, Germany); GSI, Discovery CT750 HD (GE Healthcare, Milwaukee, WI, USA); SDCT, IQon (Philips Healthcare, Best, The Netherlands). Images were acquired at the level of the left main coronary artery (LM), ascending aorta (AA) and descending aorta (DA) (Fig. 1c). For each system, a reference conventional CT acquisition was made at 120 kVp and 150 mAs. Subsequently scans were acquired using the lowest possible tube potential of 70 kVp (DSCT) and 80 kVp (GSI and SDCT), with a target tube exposure of 300 mAs. Thereafter, scans were performed in DECT modes of 80/150Sn kVp (DSCT), 80/140 kVp (GSI) and 120 kVp (SDCT) with tube exposures of 255/128 (DSCT), 160/133 (GSI) and 150 mAs (SDCT), respectively. For each acquisition, a spiral thoracic aorta protocol in the arterial phase was selected. For each CT system, a 40 ml test bolus of routine iodine concentration (300 mg I/ml) with the same fixed injection protocol was used to determine the scan delay. No bolus-tracking protocol was used because the attenuation of iodine decreases with the reduction of iodine concentration, therefore we expected that a fixed threshold (e.g. 100 HU) could not be reached by all iodine concentrations. For both DSCT and GSI, images were reconstructed using FBP. Since FBP is no longer available with SDCT, the lowest level (0) of iterative reconstruction (IR) was used for image reconstruction. Care should be taken when evaluating protocols with different reconstruction methods (IR/FBP), therefore only different scan protocols within a vendor were compared in the current study and no direct comparison between vendors was performed. Detailed acquisition parameters for all systems are listed in Table 1.

Image analysis

Images were analyzed using vendor specific dedicated workstations (SyngoVia version VB10A, Siemens Healthineers; Volume Viewer version 12.3, Ext. 4, GE Healthcare;

Table 1 Scan parameters

CT parameters	DSCT			GSI			SDCT		
Scan mode	Conv. reference	Conv. low kVp	DECT	Conv. reference	Conv. low kVp	DECT	Conv. reference	Conv. low kVp	DECT
Detector collimation (mm)	96×0.6	96×0.6	64×0.6	64×0.625	64×0.625	64×0.625	64×0.625	64×0.625	64×0.625
Tube voltage (kVp)	120	70	80/150Sn	120	80	80/140	120	80	120
Tube current (mA)	360	720	714/359	370	700	450/375	82	164	82
Gantry rotation time (s)	0.25	0.25	0.25	0.4	0.4	0.7	0.33	0.33	0.33
Exposure (mAs)	150	300	255/128	150	286	160/133	150	300	150
CTDI$_{vol}$ (mGy)	10.0	3.5	9.2	10.5	6.7	10.8	13.5	9.0	13.5
Scan delay (s) post injection	19	19	19	19	19	19	19	19	19
Pitch	0.60	0.60	0.70	0.98	0.98	0.98	0.18	0.18	0.18
Mode	Spiral	Spiral	Spiral	Spiral	Spiral	Spiral	Spiral	Spiral	Spiral
Kernel	Cardiac (Bv36d)	Cardiac (Bv36d)	Cardiac (Bv36d)	Standard	Standard	Standard	Cardiac (CB)	Cardiac (CB)	Cardiac (CB)
Slice thickness/increment (mm)	1/1	1/1	1/1	1/1	1/1	1/1	1/1	1/1	1/1
Reconstruction algorithm	FBP	FBP	FBP	FBP	FBP	FBP	iDose level 0	iDose level 0	Spectral level 0

CTDI$_{vol}$ volumetric CT dose index, *Conv.* conventional, *DECT* dual-energy CT, *DSCT* dual source CT, *FBP* filtered back projection, *GSI* gemstone spectral imaging, *kVp* kilovoltage peak, *mAs* milliampere second, *mGy* milligray, *SDCT* dual-layer spectral detector CT
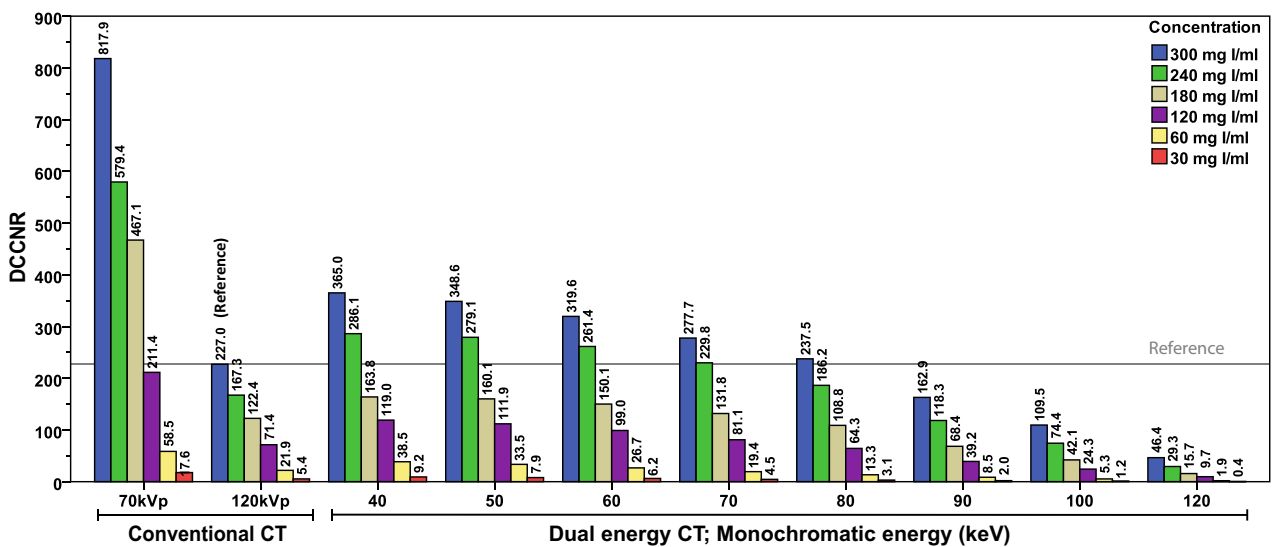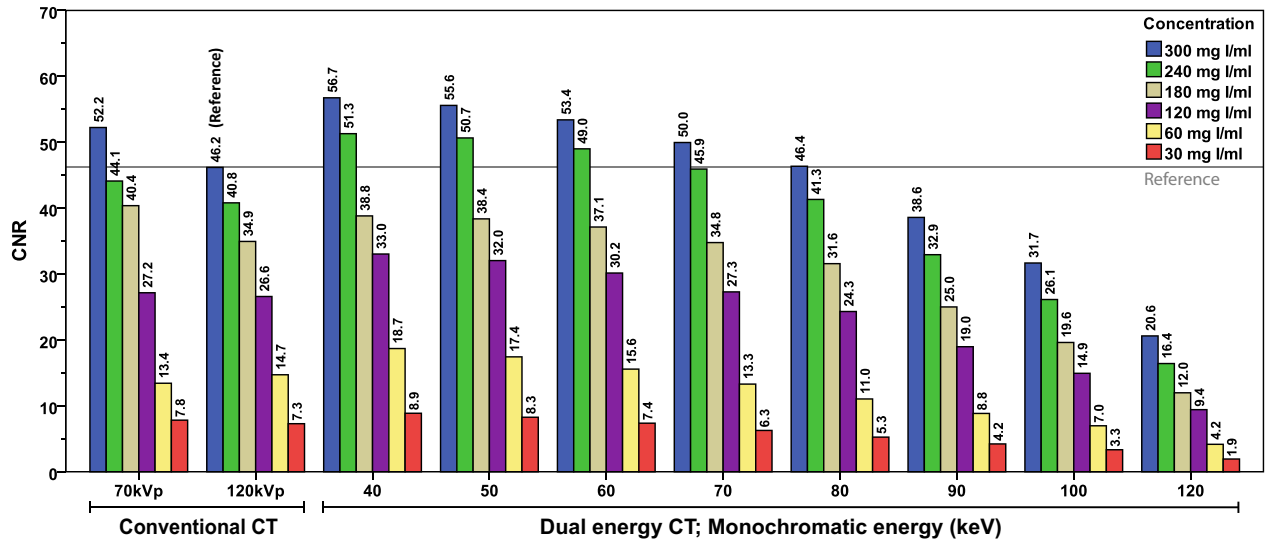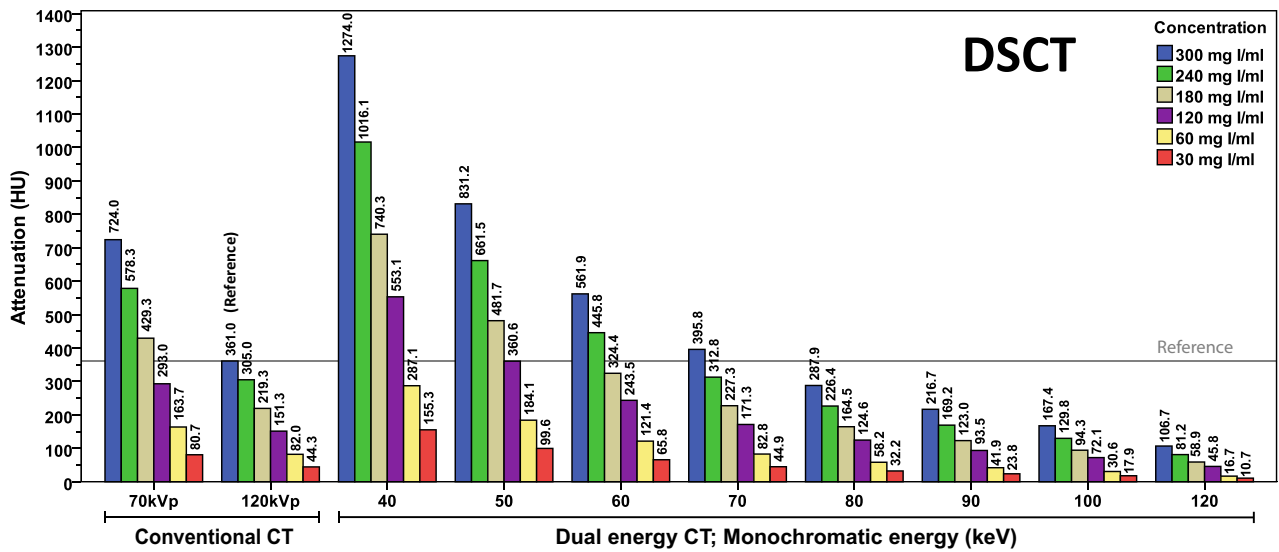
◀**Fig. 2** Objective image quality of dual source computed tomography. Average attenuation, CNR and DCCNR values per scan across different energy levels and different iodine concentrations. Reference line indicates attenuation values of routine iodine concentration (300 mg I/ml) scanned on reference conventional CT (120 kVp). Corresponding kVp, mAs and CTDI$_{vol}$ for the different scan protocols were: 70 kVp, 300 mAs, 3.5 mGy; 120 kVp, 150 mAs, 10 mGy; 80/150Sn kVp, 255/128 mAs, 9.2 mGy (DECT). *CNR* contrast to noise ratio, *CTDI$_{vol}$* volumetric CT dose index, *DCCNR* dose corrected contrast to noise ratio, *DECT* dual-energy computed tomography, *DSCT* dual source computed tomography, *HU* Hounsfield units, *kVp* kilovoltage peak, *keV* kilo electron voltage, *mAs* milliampere second, *mGy* milligray

IntelliSpace Portal version 6.5.0.02080, Philips Healthcare). DECT data were reconstructed to obtain VMI at following energy levels; 40, 50, 60, 70, 80, 90, 100 and 120 keV. On both conventional CT images and VMI (DECT), intraluminal enhancement (Hounsfield units (HU)) and noise (standard deviation (SD) of the HU measurement) were measured by manually placing a circular region of interest (ROI) with a fixed area of 0.2 (LM), 5.0 (AA) and 2.0 (DA) cm^2. Background enhancement and noise were measured by placing an ROI (4.0 cm^2) in 2 homogeneous areas in the water container (Fig. 1c). For each location, the shape, position and size of the ROI were kept consistent for all measurements. For each reconstruction, the two measurements of background enhancement and noise were averaged for further analysis of contrast to noise ratios (CNR), which was calculated as follows [19]:

$$CNR = \frac{\left(HU_{Artery} - HU_{Background}\right)}{\sqrt{\left(\frac{1}{2} \times \left(\left(SD_{Artery}{}^2\right) + \left(SD_{Background}{}^2\right)\right)\right)}}$$

Radiation dose differed between scan protocols (Table 1). To allow for evaluation of the effect of different tube voltages and VMI energies on CNR independent of dose, CNR was normalized to volumetric CT dose index (CTDI$_{vol}$) for each protocol by using dose corrected CNR (DCCNR), defined as [20–22]:

$$DCCNR = \frac{(CNR^2)}{(CTDI_{vol})}$$

Statistical analysis

The primary study measures were attenuation (HU), CNR and DCCNR. Per acquisition, measurements obtained in the 3 vessels were averaged. For each vendor, image analysis measurements of routine iodine concentration (300 mg I/ml) scanned at 120 kVp were used as a reference. Measurements with conventional CT at lowest possible kVp and DECT at different iodine concentrations were compared to the reference measurement for each CT system.

Results

## Dual source computed tomography (Fig. 2)

At reference (120kVp; 300 mg I/ml) average attenuation, CNR and DCCNR were 361.0 HU, 46.2 and 227.0, respectively. With lowest tube voltage of 70 kVp, attenuation and DCCNR increased and iodine concentration reduction down to 180 mg I/ml (40% reduced) resulted in higher attenuation (429.3 HU) and DCCNR (467.1) in comparison to the reference. For CNR at 70 kVp a small increase was observed, and dropped below the reference CNR at iodine concentration reduction of 20% or more. In DECT mode, the optimal VMI energy level with the highest attenuation, CNR and DCCNR values was 40 keV. At this keV, attenuation of reduced iodine concentration down to 120 mg I/ml (60% reduced) was 553.1 HU, and was higher compared to reference. At this same keV, CNR and DCCNR were 52.3 and 286.1 at 240 mg I/ml (20% reduced), and were higher compared to the reference. Further iodine concentration reductions resulted in attenuation, CNR and DCCNR values below the reference levels.

Gemstone spectral imaging (Fig. 3)

At reference (120kVp; 300 mg I/ml) average attenuation, CNR and DCCNR were 327.1 HU, 20.3 and 39.7, respectively. At the lowest tube voltage (80 kVp), iodine concentration reduction down to 180 mg I/ml (40% reduced) resulted in an attenuation of 354.4 HU and a DCCNR of 46.1, both higher compared to the reference. For CNR at 80 kVp, iodine concentration reduction down to 20% resulted in a higher CNR compared to the reference. With VMI on DECT, the highest attenuation was achieved at 40 keV, whilst the highest CNR and DCCNR values were obtained at 70 keV. On 40 keV images, the average CT-value of reduced iodine concentration down to 120 mg I/ml (60% reduced) was 369.6 HU, and was higher compared to reference. CNR and DCCNR values at routine iodine concentration (300 mg I/ml) on 70 keV images, were 18.9 and 33.5 and were lower compared to imaging at reference. Any (further) reduction in iodine concentration led to lower values than at reference for both conventional CT low kVp and DECT.

Dual-layer spectral detector computed tomography (Fig. 4)

At reference (120kVp; 300 mg I/ml) average attenuation, CNR and DCCNR were 311.8 HU, 28.2 and 60.2, respectively. Scanning with the lowest tube voltage of 80 kVp resulted in higher attenuation values, whereby an iodine concentration reduction of 20% resulted in 419.6 HU and
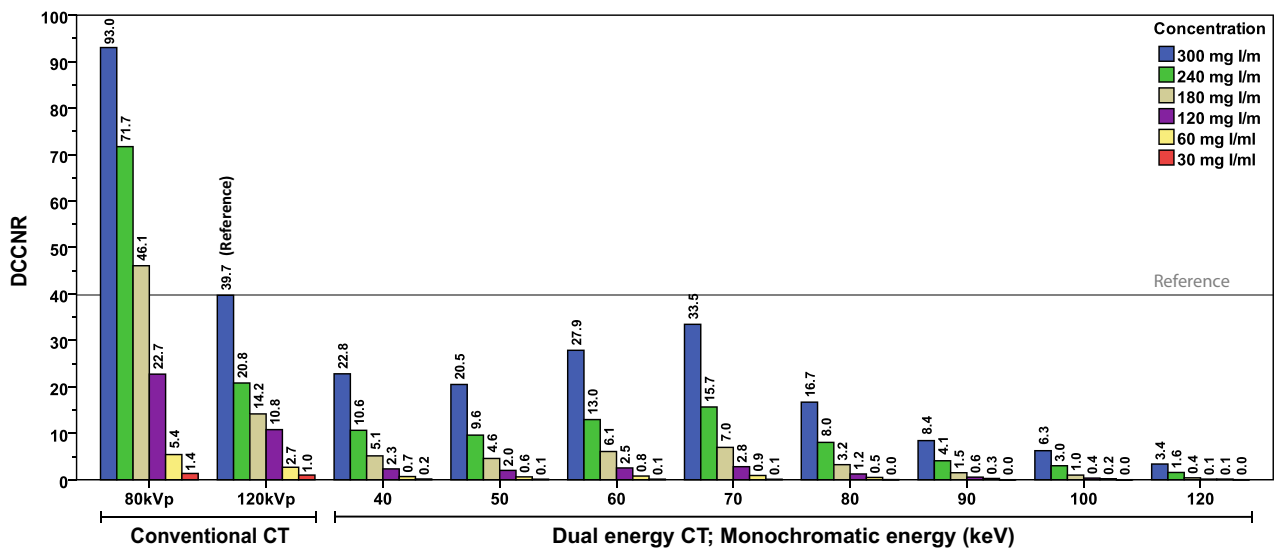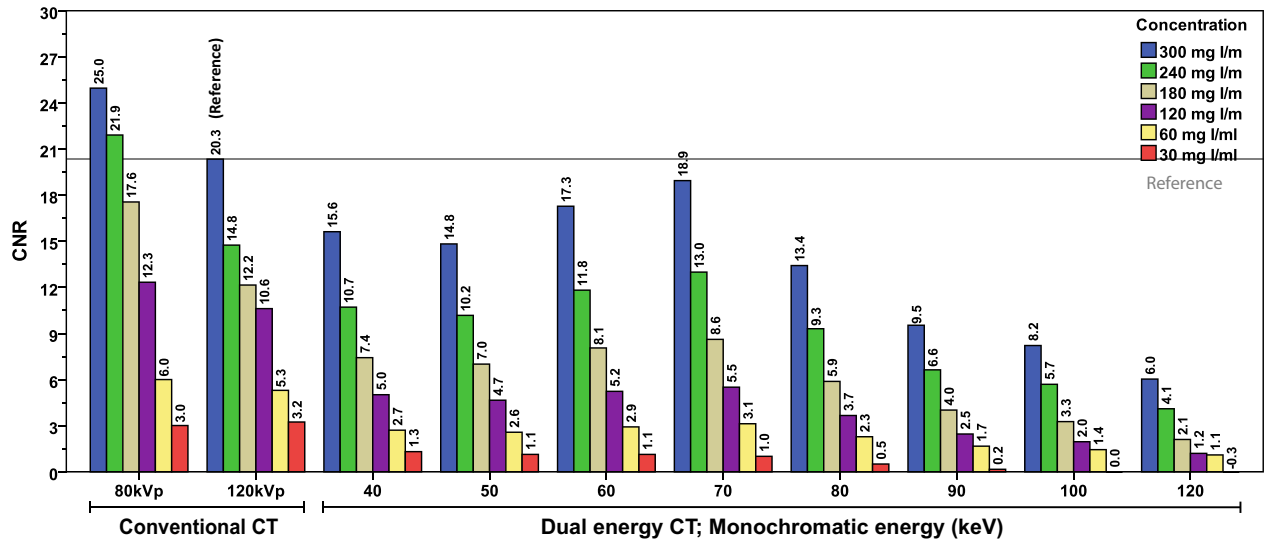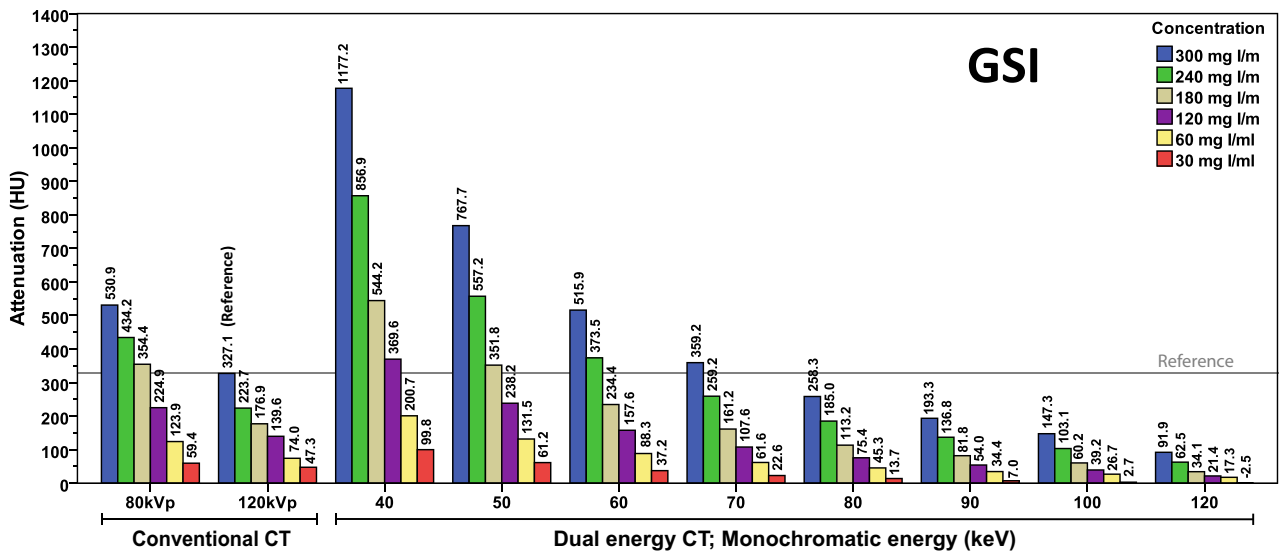
◀**Fig. 3** Objective image quality of gemstone spectral imaging. Average attenuation, CNR and DCCNR values per scan across different energy levels and different iodine concentrations. Reference line indicates attenuation values of routine iodine concentration (300 mg I/ml) scanned on reference conventional CT (120 kVp). Corresponding kVp, mAs and $CTDI_{vol}$ for the different scan protocols were: 80 kVp, 286 mAs, 6.7 mGy; 120 kVp, 150 mAs, 10.5 mGy; 80/140 kVp, 160/133 mAs, 10.8 mGy (DECT). *CNR* contrast to noise ratio, $CTDI_{vol}$ volumetric CT dose index, *DCCNR* dose corrected contrast to noise ratio, *DECT* dual-energy computed tomography *GSI* gemstone spectral imaging, *HU* Hounsfield units, *kVp* kilovoltage peak, *keV* kilo electron voltage, *mAs* milliampere second, *mGy* milligray

was higher compared to the reference. At 80 kVp, no iodine concentration reduction without loss of CNR or DCCNR compared to the reference was feasible. In DECT mode, optimal VMI energy level was 40 keV. At this energy level, iodine concentration reduction down to 120 mg I/ml (60% reduced) resulted in a higher attenuation (447.1 HU), CNR (46.9) and DCCNR (171.5) compared to the reference. Further iodine concentration reductions led to lower values compared to the reference.

Discussion

In this phantom study, we showed that for CTA images of the aorta and coronary arteries iodine reductions without loss in objective image quality compared to reference are feasible with both conventional CT low kVp and DECT. The optimal scan protocol with the highest iodine concentration reduction differed per CT system. The DCCNR results showed that an iodine concentration reduction of 40% can be achieved using conventional CT low kVp on DSCT and GSI, and 60% using DECT on SDCT. DCCNR allows for an evaluation of objective image quality independent of the effect of radiation dose, and thereby for an evaluation of the extent to which iodine concentration could be reduced purely based on tube voltage settings and VMI energy levels [20–22]. The increase in DCCNR for the low kVp protocols implicates that a tradeoff should be made between reduction of radiation dose and iodine concentration when optimizing these protocols.

With the use of a dynamic circulation phantom we were able to evaluate the effect of iodine concentration reduction on objective image quality using different CT systems while mimicking a clinical situation as best as possible by carefully controlling parameters such as heart rate, blood pressure and cardiac output. We found that each CT system has its own optimal scan protocol at which the highest iodine concentration reduction can be achieved without loss of objective image quality, when compared to the reference protocol performed on the same scanner (120kVp; 300 mg I/ml). Moreover, there were substantial differences in attenuation values, CNR and DCCNR between CT systems. An explanation for

these differences can probably be found in radiation dose, image acquisition technique (e.g. kVp, pitch, rotation time), reconstruction model (FBP and IR) and technique (kernel, VMI formation and post processing noise reduction). However, to which extent each factor played a role in the differences we found remains unclear and is beyond the scope of this study. Therefore, no direct comparison between vendors was performed.

For all scanners in DECT mode, attenuation values of iodine increased with lowering VMI energies, with highest attenuation found at 40 keV. CNR values however, did not follow the same trend for all systems (Figs. 2, 3, and 4). A different optimal keV with highest CNR values was found for GSI (70 keV) compared to DSCT and SDCT (40 keV). Because CNR is a quotient of both attenuation and noise, an explanation for the difference in CNR can be found in the noise. In VMI, noise increases for both low and high energies [23]. Therefore, each vendor has devised their own solution for this problem. One of these solutions is post processing noise reduction algorithms. GSI uses noise reduction based on IR and known noise properties from the projection based images [16]. DSCT uses an advanced image-based post processing technique (Mono+) to decrease noise on low keV [24, 25]. SDCT uses a post processing anti-correlated noise reduction algorithm which is performed on projection-based data [16, 26]. Other factors that could influence noise and thereby CNR and DCCNR, include, but are not limited to; tube voltage, tube current, pitch, kernel and image reconstruction. For image reconstruction FBP was used for DSCT and GSI and the lowest level (0) of IR for SDCT, because FBP is no longer available on this platform. However, as IR significantly reduces image noise [19], this could influence both the optimal keV selection and the CNR and DCCNR results [12, 14, 27, 28]. In addition, it is expected that when applying (higher levels of) IR, even further iodine concentration reduction could be achieved without loss of objective image quality in terms of CNR and DCCNR values for both conventional CT low kVp scans and DECT low VMI energies [7, 9, 25, 29].

Multiple studies have investigated iodine reduction using conventional CT low kVp [6–9] as well as DECT protocols [11–15, 30]. Studies on DSCT concluded that both 70 kVp and DECT allowed for iodine reduction of 44–56% [6, 8, 15] without loss of attenuation compared to a reference 120 kVp scan. For 70 kVp, CNR values were found to be significantly decreased [6] and to be similar [8] compared to reference CNR. The difference in CNR found by these studies can be explained by the fact that Thor et al. [8] used equal $CTDI_{vol}$ values for all protocols by increasing tube current for low kVp scans, whereas Kok et al. [6] did not use equal $CTDI_{vol}$. Equal $CTDI_{vol}$ is comparable to DCCNR used in our study. Findings of both studies are in accordance with our results on the DSCT; our DCCNR results are
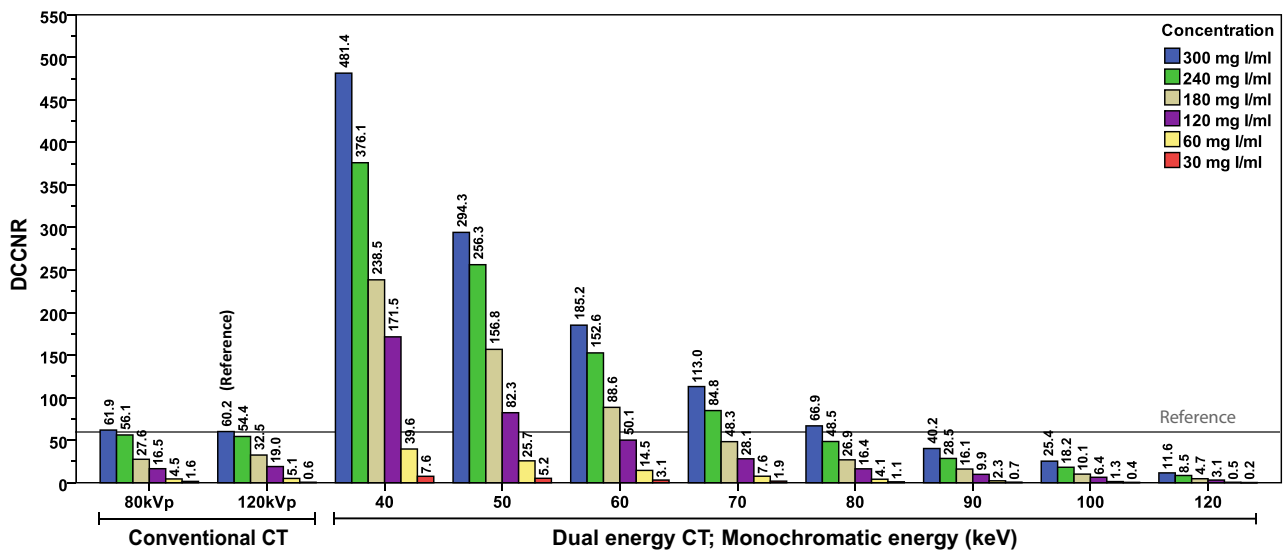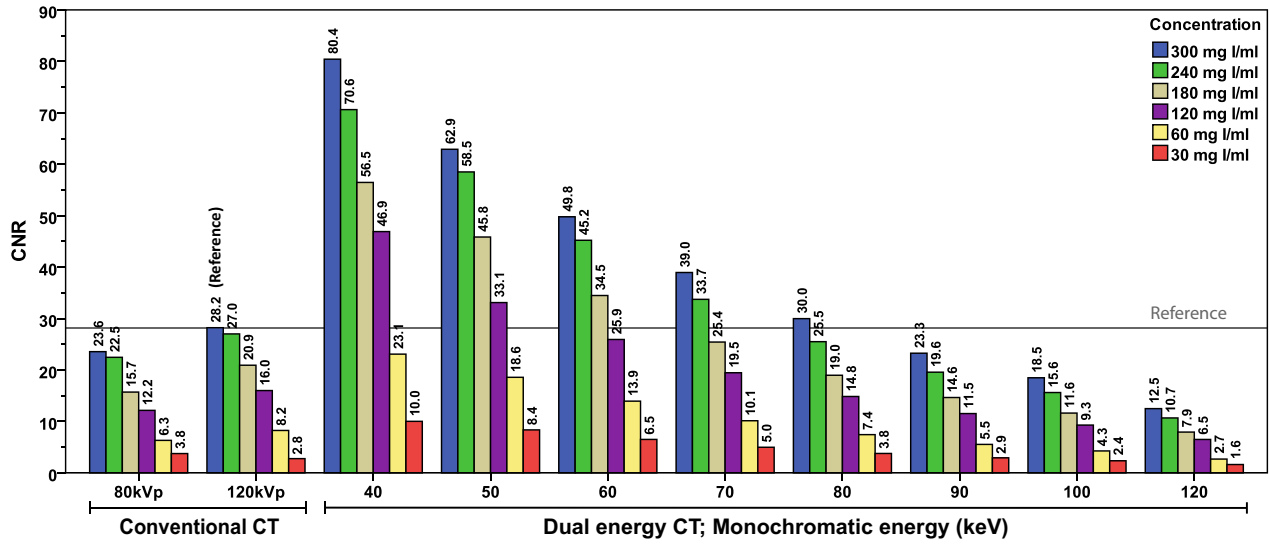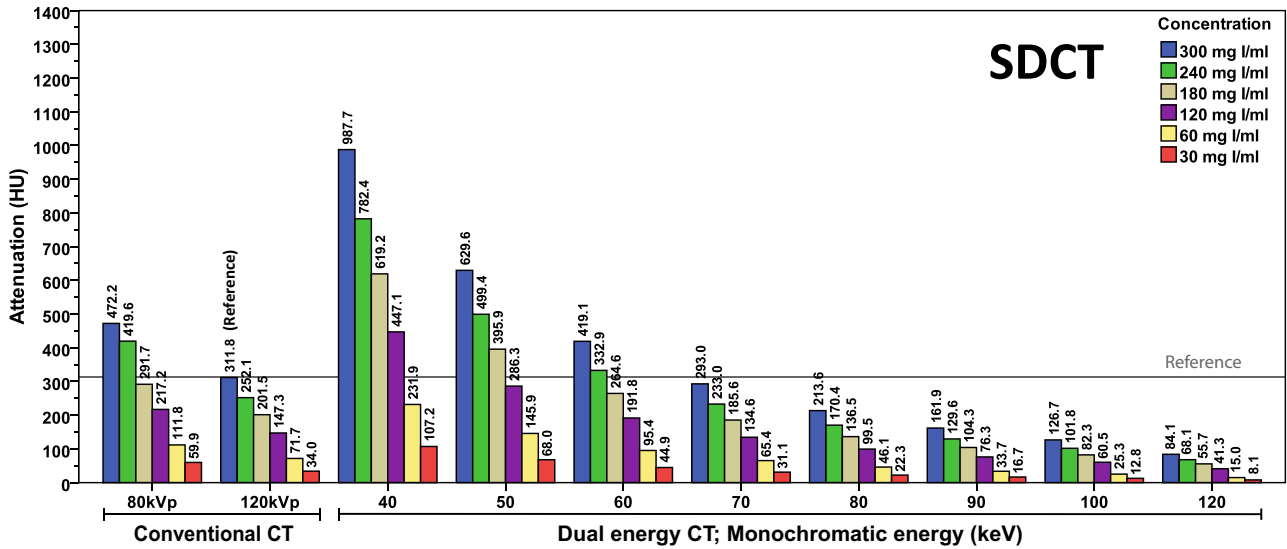
◄**Fig. 4** Objective image quality of dual-layer spectral detector computed tomography. Average attenuation, CNR and DCCNR values per scan across different energy levels and different iodine concentrations. Reference line indicates attenuation values of routine iodine concentration (300 mg I/ml) scanned on reference conventional CT (120 kVp). Corresponding kVp, mAs and CTDI$_{vol}$ for the different scan protocols were: 80 kVp, 300 mAs, 9.0 mGy; 120 kVp, 150 mAs, 13.5 mGy; 120 kVp, 150 mAs, 13.5 mGy (DECT). *CNR* contrast to noise ratio, *CTDI$_{vol}$* volumetric CT dose index, *DCCNR* dose corrected contrast to noise ratio, *DECT* dual-energy computed tomography, *HU* Hounsfield units, *kVp* kilovoltage peak, *keV* kilo electron voltage, *mAs* milliampere second, *mGy* milligray, *SDCT* dual-layer spectral detector computed tomography

in accordance with findings of Thor et al. [8] and our CNR results are in accordance with Kok et al. [6]. A study on iodine reduction with DSCT in DECT mode showed that with 60 keV images at 44% reduced iodine dose, attenuation values were similar and CNR values were significantly lower compared to reference [15]. We found similar results with DSCT in DECT mode; at 40 keV images, iodine reduction down to 60% was feasible without loss of attenuation, while for CNR more than 20% reduction led to decreased CNR values compared to reference. The discrepancy between optimal keV found in study by Delesalle et al. [15] (60 keV) and our study (40 keV) can possibly be explained by the fact that an advanced image-based post processing technique (mono+) was used in our study [15, 24, 25]. Studies on GSI concluded that 80 kVp allowed for iodine reduction down to 47% (170 mg I/ml) [7] and VMI at low energy with DECT down to 70% [11–14] without loss of attenuation and CNR. In terms of attenuation, we found similar results on GSI, with feasibility of contrast reduction down to 60%, while for CNR we found that less to no reduction was feasible. This discrepancy can possibly be explained by the usage of IR in these studies, which was not used in the current study [7, 11–14]. A recent study on SDCT showed the feasibility of iodine concentration reduction in a static phantom [30]. When compared to a reference CNR of around 30 (comparable to our reference CNR), the iodine concentration could be reduced with around 65% using low VMI at 40 keV without loss of objective image quality [30]. These results are in line with our findings of iodine contrast reduction down to 60% using low VMI at 40 keV. In another study, iodine concentration reduction using conventional CT low kVp (80 kVp) has been described on a CT system from the same vendor showing increased attenuation and decreased CNR compared to reference, similar to our study [9].

When considering optimal iodine concentration reduction, multiple factors have to be taken into account. This includes the potential to reduce radiation dose when scans are made using conventional CT low kVp. However, low kVp imaging may not be the preferred strategy for imaging larger patients. On the other hand, DECT provides extra information such as iodine density, which is a semi-quantitative indicator for perfusion [31].

Although the dynamic circulation phantom setup allowed for an evaluation of optimal iodine reduction in a controlled setting mimicking a clinical situation across multiple vendors, it has limitations. First, the phantom is relatively small, and the results of the present study may therefore be limited to small to average-sized patients. In addition, physiological parameters, such as cardiac output, can differ per patient. Second, as described before, IR was only selected for SDCT image reconstructions, because no FBP was available on this scanner. Even though the lowest level of IR was used, this could still have influenced our CNR and DCCNR results by decreasing noise for this vendor. For this reason, we did not make a direct comparison between vendors and only compared results of different scan protocols within one vendor. Third, no subjective image quality was evaluated. Patient studies on both subjective and objective image quality are needed to confirm our findings in routine clinical care, whereby results of the current study could be used as a guideline for protocol development on a per vendor level. Fourth, in the current study a fixed injection protocol was used whereas in clinical practice a multi-stage injection method is more commonly used. The fixed injection protocol was chosen to keep iodine administration similar between the different CT systems.

In conclusion, we demonstrated that iodine concentration reduction is feasible with both low kVp on conventional CT and low VMI energies on DECT without loss of objective image quality. The optimal scan protocols at which highest iodine concentration reduction can be achieved differed per vendor. Maximal iodine concentration reduction of up to 40% is feasible when using conventional CT low kVp on DSCT (70 kVp) and GSI (80 kVp), and up to 60% when using DECT (40 keV) on SDCT without loss of objective image quality compared to routine iodine concentration imaged at 120 kVp. However, patient studies are needed to extend and translate our findings.

Compliance with ethical standards

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