Feasibility of multi-contrast imaging on dual-source photon counting detector (PCD) CT: An initial phantom study

Shengzhen Tao, Kishore Rajendran, Cynthia H. McCollough, and Shuai Leng^{a)}
Department of Radiology, Mayo Clinic, Rochester, MN, USA

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Purpose: Photon-counting-detector-computed tomography (PCD-CT) allows separation of multiple, simultaneously imaged contrast agents, such as iodine (I), gadolinium (Gd), and bismuth (Bi). However, PCDs suffer from several technical limitations such as charge sharing, K-edge escape, and pulse pile-up, which compromise spectral separation of multi-energy data and degrade multi-contrast imaging performance. The purpose of this work was to determine the performance of a dual-source (DS) PCD-CT relative to a single-source (SS) PCD-CT for the separation of simultaneously imaged I, Gd, and Bi contrast agents.

Methods: Phantom experiments were performed using a research whole-body PCD-CT and head/abdomen-sized phantoms containing vials of different I, Gd, Bi concentrations. To emulate a DS-PCD-CT, the phantoms were scanned twice on the SS-PCD-CT using different tube potentials for each scan. A tube potential of 80 kV (energy thresholds = 25/50 keV) was used for low-energy tube, while the high-energy tube used Sn140 kV (Sn indicates tin filter) and thresholds of 25/90 keV. The same phantoms were scanned also on the SS-PCD-CT using the chess acquisition mode. In chess mode, the 4×4 subpixels within a macro detector pixel are split into two sets based on a chess-board pattern. With each subpixel set having two energy thresholds, chess mode allows four energy-bin data sets, which permits simultaneous multi-contrast imaging. Because of this design, only 50% area of each detector pixel is configured to receive photons of a pre-defined threshold, leading to 50% dose utilization efficiency. To compensate for this dose inefficiency, the radiation dose for this scan was doubled compared to DS-PCD-CT. A 140 kV tube potential and thresholds = 25/50/75/90 keV were used. These settings were determined based on the K-edges of Gd, and Bi, and were found to yield good differentiation of I/Gd/Bi based on phantom experiments and other literature. The energy-bin images obtained from each scan (scan pair) were used to generate I-, Gd-, Bi-specific image via material decomposition. Root-mean-square-error (RMSE) between the known and measured concentrations was calculated for each scenario. A 20-cm water cylinder phantom was scanned on both systems, which was used for evaluating the magnitude of noise, and noise power spectra (NPS) of I/Gd/Bi-specific images.

Results: Phantom results showed that DS-PCD-CT reduced noise in material-specific images for both head and body phantoms compared to SS-PCD-CT. The noise level of SS-PCD was reduced from 2.55 to 0.90 mg/mL (I), 1.97 to 0.78 mg/mL (Gd), and 0.85 to 0.74 mg/mL (Bi) using DS-PCD. NPS analysis showed that the noise texture of images acquired on both systems is similar. For the body phantom, the RMSE for SS-PCD-CT was reduced relative to DS-PCD-CT from 10.52 to 2.76 mg/mL (I), 7.90 to 2.01 mg/mL (Gd), and 1.91 to 1.16 mg/mL (Bi). A similar trend was observed for the head phantom: RMSE reduced from 2.59 (SS-PCD) to 0.72 (DS-PCD) mg/mL (I), 2.02 to 0.58 mg/mL (Gd), and 0.85 to 0.57 mg/mL (Bi).

Conclusion: We demonstrate the feasibility of performing simultaneous imaging of I, Gd, and Bi materials on DS-PCD-CT. Under the condition without cross scattering, DS-PCD reduced the RMSE for quantification of material concentration in relative to a SS-PCD-CT system using chess mode. © 2019 American Association of Physicists in Medicine [<https://doi.org/10.1002/mp.13668>]

Key words: dual-source CT, multi-contrast imaging, multi-energy CT, photon counting detector, spectral CT

1. INTRODUCTION

The scintillating x-ray detectors used in clinical computed tomography (CT) systems integrate the absorbed energy from all detected photons. Multi-energy acquisitions, such as required for dual-energy (DE) CT, are accomplished by a number of approaches, including those that use two tube

potentials, that is, dual-source (DS) CT or fast kV switching, and those that use a single tube potential, that is, split-beam-filter or dual-layer-detector methods.¹ Photon counting detector (PCD) CT is an emerging technology that has recently been used to image human subjects.^{2–5} Compared to conventional energy-integrating-detector (EID) technology, PCDs have demonstrated several advantages from their direct

conversion of x-ray photons to electronic signals, such as reduced electronic noise due to energy thresholding,⁶ improved iodine contrast by removing the energy weighting in the detected signals,^{7,8} and dose-efficient high-resolution imaging.⁹ Another unique feature of PCD-CT is its capability to discriminate the energy of incident photons, which makes it intrinsically suitable for multi-energy CT applications.¹⁰ With PCD-CT, a series of energy-bin images can be generated from a single data acquisition, with each energy-bin image representing the photons detected within a user-defined energy window. This allows PCD-CT to measure the change in x-ray attenuation due to the K-edges present in materials such as gadolinium (Gd) and bismuth (Bi).^{11–20}

This energy-discriminating capability of PCD-CT facilitates the separation of simultaneously imaged contrast materials. This is especially relevant for imaging of novel nanoparticle-based contrast agents having a detectable K-edge in the diagnostic x-ray energy range.^{21–24} After tagging them with relevant molecules, these nanoparticles can be used as probes to selectively image a specific target or pathology of interest, such as a vulnerable plaque. Multiple nanoparticles (e.g., iodine (I), Gd and Bi) can be administered to simultaneously image different targets or molecular pathways of interest. The use of these targeted contrast agents may extend the frontier of CT beyond conventional anatomical imaging toward molecular imaging to eventually enable novel CT applications. In addition, the use of multiple contrast agents may be relevant for clinical applications that require multi-phase acquisitions,^{11,14,19,20,25–27} where repeated scans are required to capture the contrast enhancement at different time points, which can lead to increased radiation dose.²⁸ With multi-contrast imaging, a single scan is acquired after injection of different contrast agents at different time points. Then, a series of contrast-specific images can be generated retrospectively, using material decomposition, to depict each contrast enhancement phase.^{11,22,29,30} The use of multiple contrast agents can therefore reduce the number of scans and might present an opportunity to reduce patient radiation dose.

The performance of multi-contrast imaging relies on the degree to which the acquired multi-energy datasets represent unique distributions of photon energies. Limitations of PCD-CT technology, such as K-escape, charge sharing, and pulse pileup, cause the energy response of each energy bin to deviate considerably from the ideal response.^{31–34} The resulting imperfect separation of the acquired energy distributions (i.e., the presence of spectral overlap) degrades the performance of multi-contrast imaging. With EID-based CT, DE datasets can be acquired by using two sets of tube/detector pairs (i.e., DSCT), with each pair operating at different tube potentials.¹ An additional beam filter, such as tin, can be added to one of x-ray tube to improve spectral separation.³⁵ The same approach can be utilized in PCD-CT, yielding a DS-PCD-CT system architecture. Previous works based on numerical simulation and phantom experiments have shown that such a DS-PCD-CT may provide benefit for conventional DE imaging applications such as iodine- and water-based material

decomposition.^{36,37} Compared to two-material decomposition, multi-contrast imaging is a more challenging task with images more susceptible to imperfect PCD spectral response, as well as noise amplification during material decomposition. By combining DS architecture and PCD technology, the spectral separation among the acquired energy-bin datasets may be improved, which can potentially benefit multi-contrast imaging. The purpose of this work was to demonstrate the feasibility of performing multi-contrast imaging of I, Gd, and Bi contrast agents on a DS-PCD-CT platform based on phantom experiments, and compare its performance to an existing single-source PCD-CT using chess acquisition mode. A DS-PCD-CT was emulated in this study by scanning the same object twice with different tube potentials due to the absence of a true DS-PCD system. Because of this setup, the effect of cross scattering between the two subsystems will not be included. The cross scattering can affect the spectral separation of the data acquired on a DS-PCD system. However, the dual-scan setup still allows us to demonstrate the key concept and to answer the question regarding the performance of DS-PCD using realistic PCD units under the condition without cross scattering. This is the first step toward the development and investigation of such systems and has important implication and meaning for showing the benefits of DS-PCD system.

2. MATERIALS AND METHODS

2.A. Research Whole-Body PCD-CT

Phantom experiments involving three contrast materials including I, Gd, and Bi were performed on a research whole-body PCD-CT (Somatom CounT, Siemens Healthcare GmbH, Forchheim, Germany).⁸ This system is a second generation DSCT (Somatom Definition Flash, Siemens Healthcare GmbH), where the detector array on one of the two tube/detector pairs was replaced with a PCD array. The x-ray source associated with the PCD array is the same as that associated with the EID array. Additionally, a 0.4 mm tin (Sn) filter can be inserted into the x-ray beam of the tube associated with the PCD array. A true DS-PCD-CT system architecture is *emulated* on this single-source (SS) PCD-CT by scanning the same phantom twice using two different tube potential/beam filter combinations.

Currently, four data acquisition modes have been implemented on the SS-PCD-CT system that is, macro, chess, sharp, and ultra-high resolution modes.⁹ Different acquisition modes have different numbers of energy thresholds, minimum detector readout pixel sizes, and longitudinal collimations. For the purpose of this work, we focus on the macro and chess mode, which have the same detector pixel size at isocenter (0.5 mm) and the same longitudinal coverage of $32 \times 0.5 = 16$ mm. The macro mode allows two energy thresholds, which can be used to generate two unique energy-bin datasets. The chess mode splits a detector pixel into two sets of interleaved subpixels based on a chess-board pattern. With each of the two subpixels sets

having two energy thresholds, the chess mode allows a total of four unique energy-bin data sets.⁸ It is therefore capable of multi-contrast imaging using I, Gd, and Bi. However, due to the chess-board pattern design, only half of the detector pixel area in chess mode is configured to receive photons of a pre-defined threshold, leading to 50% dose utilization efficiency. As a result, the radiation dose applied in the chess mode should be doubled to get the same dose being used for image formation as a true PCD with four energy thresholds. Note that the chess mode would not be used clinically and was created on this research system simply to allow acquisition of four energy bins in a single scan with two counters for each subpixel. A detector with four energy thresholds in every detector pixel would not suffer from this dose penalty; however, the spectral overlap between each energy bin would be similar. Charge sharing effect for the chess mode used here is equivalent to a true 4-threshold detector despite their difference in readout electronic setup. This is because charge sharing happens before the readout electronics. Each detector subpixel will receive the charge shared from neighboring subpixels, with the amount of charge solely determined by physical interaction and spatial cross talk.

2.B. Experiments

Numerical simulation was first performed to compare the spectral separation between the SS-PCD-CT chess mode and an *emulated* DS-PCD-CT macro mode using vendor-provided simulation tool developed for the research whole-body PCD-CT system.³⁸ The simulation tool can model the deposition of x-ray energy in the direct converting material (e.g., CdTe) due to various physical interactions including Rayleigh scatter, Compton scatter, and photoelectric effect. It can also simulate the propagation of the charge clouds and consider imperfect detector responses due to charge sharing, fluorescence x-rays, and cross-talk between different detector pixels, as well as simulate the effect of energy thresholding and electronic noise.³⁹ As such, the energy spectra of transmitted x-ray beam were generated using the simulation tool after accounting for both input x-ray spectrum and detector response. SS-PCD-CT chess mode acquisition was performed with 140 kV tube potential to provide the widest range of energy thresholds, which were set as 25, 50, 75, 90 keV, yielding a total of four energy-bin datasets. The 25 keV threshold was used to reject electronic noise, while the 50 and 90 keV thresholds were used to capture the K-edge of Gd and Bi, respectively. Setting thresholds based on K-edges helps improving the differentiation of these contrast agents, as shown in various studies.^{13,14,22} The 75 keV threshold was selected to balance the number of photons within adjacent energy bins. Based on previous phantom experiments and other literature,¹¹ these thresholds yield good performance for differentiating I, Gd, and Bi contrast agents. The 140 kV tube potential allows more photons to be included in the highest energy bin compared to a lower tube potential such as 120 kV.

For the *emulated* DS-PCD-CT, tube potentials of 80 and 140 kV were used to maximize the energy separation between the low- and high-energy scans. A 0.4 mm Sn filter was used for the 140 kV scan to further improve energy separation. Macro mode, with its two energy thresholds, was used for the *emulated* DS-PCD-CT. The energy thresholds were set at 25 and 50 keV for the 80 kV scan and 25 and 90 keV for 140 kV + Sn scan, yielding a total of four energy-bin images, the same as for the SS-PCD-CT chess mode acquisition. The detected energy spectra after passing through 20 and 30 cm diameter water phantom were generated using vendor-provided software,³⁸ which simulated the attenuation of the head and abdomen phantoms, respectively.

Phantom experiments were then performed to evaluate the performance of multi-contrast imaging using DS-PCD-CT. To emulate head and abdominal CT scans, a 20-cm diameter circular head phantom made from water-equivalent material (Gammex Inc., Middleton, WI, USA), as well as an abdominal-shaped (30 × 25 cm) water phantom, was scanned on the PCD-CT system using the tube potentials and acquisition modes described above. Vials containing I, Gd and Bi solutions of different concentrations were placed within each phantom. Iodine solutions with concentrations of 5 and 10 mg/mL were generated by diluting a clinical iodine contrast agent (Omnipaque 300, GE Healthcare, Milwaukee, WI, USA) with water. Gadolinium solutions of 4 and 8 mg/mL concentration were prepared from a clinical Gd contrast agent (Dotarem, Guerbet, Princeton, NJ, USA). Finally, 5 and 10 mg/mL Bi solutions were prepared by diluting Pepto Bismol solution (Pepto Bismol, Procter and Gamble, Cincinnati, OH, USA). In addition, two multi-contrast mixture solutions were also prepared, with one containing 5 mg/mL of I and 4 mg/mL of Gd, and the other containing 5 mg/mL of Bi, 4 mg/mL of Gd and 5 mg/mL of I.

To *emulate* a DS-PCD-CT scan, the head phantom was scanned twice on the SS-PCD-CT system in macro mode using 0.55 spiral pitch, collimation = 32×0.5 mm, rotation time = 1 s, full scan, 474 mAs at 80 kV and 360 mAs at Sn140 kV, and CTDI_{vol, 16cm} of 26.35 mGy at 80 kV and 26.45 mGy at Sn140 kV. The energy thresholds were set at 25/50 keV for the 80 kV scan and 25/90 keV for the Sn140 kV scan. The total radiation dose for the *emulated* DS-PCD-CT scan (52.8 mGy CTDI_{vol, 16cm}) was similar to that of a clinical head CT scan. The head phantom was also scanned on the SS-PCD-CT with the chess mode using 0.55 spiral pitch, collimation = 32×0.5 mm, rotation time = 1 s, full scan, 416 mAs, and a CTDI_{vol, 16cm} of 106 mGy. The tube potential and energy thresholds were the same as used in simulation. Note that, because of the 50% dose efficiency of the chess mode, the SS-PCD-CT scan was performed at twice the radiation dose compared to the DS-PCD-CT to compensate for this dose inefficiency.

Next, the abdominal water tank phantom containing I, Gd, and Bi solutions was scanned using the *emulated* DS-PCD-CT protocol (80/Sn140 kV, collimation = 32×0.5 mm, rotation time = 0.5 s, full scan, 316/200 mAs, CTDI_{vol, 32cm} = 7.0/6.95 mGy, energy thresholds = 25/50 keV and 25/90 keV for

the 80 and Sn140 scans, respectively). The total radiation dose level (13.95 mGy CTDI_{vol, 16cm}) of the DS-PCD scan was similar to that of a routine clinical abdominal CT scan. The same body phantom was scanned using the SS-PCD chess mode abdominal protocol with 140 kV, energy thresholds = 25/50/75/90 keV, collimation = 32×0.5 mm, rotation time = 0.5 s, full scan, 252 mAs, CTDI_{vol, 32cm} = 27.91 mGy. Again, the total radiation dose of the SS-PCD-CT chess mode acquisition was twice as that of DS-PCD-CT.

A 20 cm diameter cylindrical water phantom was further scanned on DS-PCD-CT and SS-PCD-CT using the same acquisition settings as the head phantom experiment to evaluate the noise power spectra. Each acquisition was repeated three times.

2.C. Data processing

For each acquisition, the energy-bin images were reconstructed with a weighted filtered back projection algorithm⁴⁰ using a medium-smooth quantitative (D30) kernel available with a vendor-provided off-line reconstruction package (Recon CT, Siemens Healthcare GmbH). Images were reconstructed with a field-of-view of 25 cm, and a 512 × 512 matrix size. The image thickness for the head and body scans were 2 and 5 mm, respectively. Next, material decomposition was performed using an image domain least-square-based approach with volume conservation constraint, assuming I, Gd, Bi, and water as basis materials. The mean values and standard deviations of material concentrations were measured for each vial using a 10-mm diameter circular region-of-interest. The root-mean-square-error (RMSE) for quantification of material concentration between the known and measured concentrations, which includes the effects of bias and noise, was also calculated for comparison between the SS-PCD chess mode scans and the *emulated* DS-PCD macro-mode scans. The RMSE calculation included all pixels within the ROIs in different contrast vials.

The noise power spectra (NPS) of each material-specific images decomposed using SS-PCD and DS-PCD images were generated from the water phantom scans. To calculate NPS, a total of 40 square ROIs (9 × 9 mm²) located on a 70 mm radius circle were extracted from the I-, Gd-, and Bi-specific images obtained from the water scan. The NPS curves of each material-specific image were then generated following the method described by Siewerdsen et al.⁴¹, and then averaged among three repeated scans. The image noise level of each material images was also calculated as the standard deviation measured using a cylindrical ROI (diameter = 40 mm, length = 10 mm) in the center of the phantom.

3. RESULTS

3.A. Simulations

Figures 1(a) and 1(c) shows the energy spectra for the four energy bins of a SS-PCD chess mode acquisition after 20 and 30 cm of water attenuation, respectively. Note that the long

high-energy tails of the lower energy-bin spectra (e.g., 25–50 and 50–75 keV bins) are due to the imperfect detector response³³ and lead to substantial overlap between different energy bins. The mean energies of the four bins are 71.8, 75.5, 92.5, 109.3 keV with 20 cm of water attenuation, and are 76.0, 78.1, 93.6, 110.1 keV with 30 cm of water attenuation. In both cases, the mean energies of the 25–50, 50–75, and 75–90 keV energy bins exceed the supposed upper energy limit of the bins, with the mean energy of 25–50 keV bin being almost equivalent to that of the 50–75 keV bin.

Figures 1(b) and 1(d) shows the energy spectra of the four energy bins for an *emulated* DS-PCD macro-mode acquisition after 20 and 30 cm of water attenuation, respectively. The mean energy of the four energy bins are 57.2, 63.9, 87.7, 110.9 keV and 59.2, 64.7, 90.3, 111.7 keV for 20 and 30 cm of water attenuation, respectively. The *emulated* DS-PCD approach substantially reduced the mean energies of the 25–50, 50–75, and 75–90 keV energy bins. For the 25–50 keV energy bin, for example, the mean energy is 71.8 keV for SS-PCD chess mode approach vs 57.2 keV using the DS-PCD macro-mode approach. The differences in mean energies between adjacent energy bins are summarized in Table I, which shows that the energy separation between adjacent energy bins increased using the DS-PCD approach.

3.B. Experiments

Figure 2 shows the material-specific images for I, Gd, and Bi in the head phantom for the SS-PCD chess mode and *emulated* DS-PCD macro-mode scan data. The low-energy threshold (TL) image of the head phantom is also shown. For the SS-PCD-CT, noise in the Bi image is lower than Gd/I images. This is due to better energy separation in higher energy photon range. As shown in Table I, the difference of mean energy between bins 1 and 2 is 3.7 keV while that between bins 3 and 4 is 16.8 keV, using SS-PCD-CT. The image noise in Gd- and I-specific images generated from DS-PCD-CT is reduced compared to that generated from SS-PCD-CT. This is in part due to the improved energy separation for the lower energy bins using the DS-PCD approach (Table I) compared to SS-PCD-CT. As shown in Fig. 3, the I, Gd, and Bi mass concentration RMSE values for the DS-PCD approach (0.72, 0.58, and 0.57 mg/mL) were reduced compared to the 2 × dose SS-PCD approach (1.93, 1.50, and 0.64 mg/mL), yielding a 63%, 61%, 10% reduction for I, Gd, and Bi mass quantification, respectively. Table II summarizes the mean values and standard deviations of material concentrations measured for each vial. Both techniques showed low quantification bias comparing between the mean concentration values and the corresponding true values, while the DS-PCD results showed consistently lower noise in different material-specific images compared to SS-PCD.

The I, Bi, and Gd material images generated from the body phantom scans performed using SS-PCD and DS-PCD are shown in Fig. 4, along with the corresponding CT (TL) image. Compared to the head mode scan in Fig. 2, the image noise in the I- and Gd-specific images of the body phantom

FIG. 1. Energy spectra of four energy bins for: (a, c) single-source (SS)-PCD chess mode scans with energy thresholds of 25, 50, 75, and 90 keV after passing through 20 cm (a) and 30 cm (c) of water attenuation, and (b, d) *emulated* dual-source (DS)-PCD macro-mode scans using 80 kV (energy thresholds = 25 and 50 keV) and 140 kV with tin filter (energy thresholds = 25 and 90 keV) for 20 cm (b) and 30 cm (d) of water attenuation. The mean energies of each bin (E_m) are provided in the figure legend. [Color figure can be viewed at wileyonlinelibrary.com]

are substantially higher, which obscured the delineation of the phantom inserts. This is in part because of the narrower difference in energy spectra of the lower energy bins (e.g., 2.1 keV between bin 1 and bin2) when imaging a larger object. Note that a larger object could further harden the x-ray beam, reduce the separation between lower energy bins, and degrade the material decomposition performance. On the other hand, the multi-contrast imaging performance is substantially improved using the DS-PCD approach. The material quantification RMSE values of images acquired from DS-PCD scan and SS-PCD scan with double radiation dose are summarized in Fig. 5. Compared to the head phantom scan, the Gd and I quantification RMSEs are increased by four times, and are 9.83, 6.84, 1.55 mg/mL for I, Gd and Bi, respectively. Using the DS-PCD approach, the RMSE values of I, Gd, and Bi quantification are reduced to 2.76, 2.01, and 1.16 mg/mL — a 71%, 71%, and 25% reduction, respectively.

The I, Gd, and Bi concentrations measured from each phantom inserts are summarized in Table III, which shows that both measurement bias and noise are reduced using DS-PCD compared to the SS-PCD chess mode, especially for the Gd and I images. More material cross-talk between Gd and I signals can be observed in the SS-PCD results compared to DS-PCD (see Mixtures 2 in Table III), as shown by the more positively biased I signal (+2.2 mg/mL for SS-PCD; +0.5 mg/mL for DS-PCD) and negatively biased Gd signal (−1.4 mg/mL for SS-PCD; −0.5 mg/mL for DS-PCD).

Figure 6 shows the NPS curves of I-, Gd-, and Bi-specific images generated from water scans performed on SS-PCD-CT and DS-PCD-CT. The NPS of the two systems have similar shapes between each other, while the amplitude of NPS is reduced using DS-PCD. This indicates that the noise texture is similar between the two systems. The noise levels in the I-, Gd-, Bi-specific images were reduced from 2.55, 1.97,

TABLE I. The differences in mean energies between adjacent energy bins using single-source PCD chess mode and dual-source PCD macro-mode acquisitions after 20 and 30 cm of water attenuation.

	Difference in mean energies of Bins 1 and 2 (keV)	Difference in mean energies of Bins 2 and 3 (keV)	Difference in mean energies of Bins 3 and 4 (keV)
20 cm (Head)			
SS-PCD	3.7	17.0	16.8
DS-PCD	6.7	23.8	23.2
30 cm (Body)			
SS-PCD	2.1	15.5	16.5
DS-PCD	5.5	25.6	21.4

0.85 mg/mL using SS-PCD to 0.90, 0.78, 0.74 mg/mL using DS-PCD.

4. DISCUSSION

In this work, we evaluated the ability of a DS-PCD-CT acquisition strategy to improve the energy separation in multi-energy CT datasets compared to a SS-PCD-CT approach that acquired four energy-bin datasets in a single scan. The DS-PCD approach was emulated using two SS-PCD scans with each acquired with different tube potential and energy threshold settings, which also provided four energy-bin datasets. The DS-PCD approach was evaluated using both head

FIG. 3. The root-mean-square-error (RMSE) of mass concentration values for iodine (I), gadolinium (Gd), and bismuth (Bi) in the head phantom images for data acquired using a single-source (SS)-PCD chess mode scan and dual-source (DS)-PCD macro-mode scans. The RMSE calculation included all pixels within the ROIs in different contrast vials. [Color figure can be viewed at wileyonlinelibrary.com]

and body phantoms. The radiation dose used in our phantom experiments was representative of routine clinical CT exams for head and body, which allowed us to evaluate the performance of DS-PCD under a clinically encountered scenario. The results clearly demonstrated that DS-PCD can improve the spectral separation and mass concentration quantification

FIG. 2. The iodine (I, a and d), gadolinium (Gd, b and e), and Bismuth (Bi, c and f) images generated from the head phantom scans using a single-source PCD chess mode acquisition (a–c) and dual-source PCD macro-mode acquisition (d–f). The CT image (energy threshold-low image) of the phantom setup is also shown. There are a total of eight vials within the phantom: two containing I (5, 10 mg/mL), two containing Gd (4, 8 mg/mL), two containing Bi (5, 10 mg/mL), as well as two containing mixtures (M1: 5 mg/mL I + 4 mg/mL Gd; M2: 5 mg/mL I + 4 mg/mL Gd + 5 mg/mL Bi). Window/level = 600/100 HU (CT) and 15/7.5 mg/mL (I, Gd, Bi images). [Color figure can be viewed at wileyonlinelibrary.com]

for I, Gd, and Bi. Because we used FBP reconstruction and least-square based material decomposition method which are linear operators, the noise of the material decomposition results acquired with different dose levels can be predicted based on an inverse square root relationship between noise and dose.

Due to various technical limitations, the multi-energy datasets acquired with SS-PCD-CT can suffer from imperfect PCD response. The imperfect PCD response causes a long tail in low energy-bin energy spectra that spreads into high-energy range, which decreases spectral separations between these datasets. As demonstrated in previous works,^{35,42} energy separation between multi-energy datasets affects image noise in the results of material

decomposition. In this work, we demonstrated decreased image noise and improved quantification accuracy using DS-PCD images. This can be explained by improved energy separation among the multiple energy-bin datasets^{35,42}. The DS-PCD reduces the high-energy tails in low energy-bin spectra by using a lower tube potential to acquire the lower energy-bin datasets. The improvement in image quality (i.e., decreased image noise) and increased quantification accuracy was more significant for I and Gd compared to Bi. This is because the K-edge of Gd is around 50 keV, and the spectral response is worst in the lower energy range. In contrast, Bi has a K-edge around 90 keV, where the spectra are not as affected by the long inaccurate high-energy tail.

TABLE II. Iodine (I), gadolinium (Gd), and bismuth (Bi) mass concentration measurements (mean ± standard deviation) for each vial in the head phantom for data acquired using a single-source (SS)-PCD chess mode scan and dual-source (DS)-PCD macro-mode scans.

	I 5 mg/mL	I 10 mg/mL	Gd 4 mg/mL	Gd 8 mg/mL	Bi 5 mg/mL	Bi 10 mg/mL	Mixture 1 (I 5 mg/mL + Gd 4 mg/mL)	Mixture 2 (I 5 mg/mL + Gd 4 mg/mL + Bi 5 mg/mL)
SS-PCD with double radiation dose (CTDI _{vol,16cm} = 106 mGy)	5.2 ± 1.9	10.0 ± 1.7	3.5 ± 1.4	8.0 ± 1.4	4.9 ± 0.6	10.1 ± 0.7	5.4 ± 2.0 (I) 3.9 ± 1.5 (Gd)	4.9 ± 2.0 (I) 4.1 ± 1.6 (Gd) 5.2 ± 0.6 (Bi)
DS-PCD (CTDI _{vol,16cm} = 53 mGy)	5.1 ± 0.7	9.9 ± 0.8	4.0 ± 0.6	8.0 ± 0.6	4.8 ± 0.5	10.1 ± 0.6	5.2 ± 0.7 (I) 4.0 ± 0.6 (Gd)	5.0 ± 0.7 (I) 3.9 ± 0.6 (Gd) 5.2 ± 0.5 (Bi)

FIG. 4. The iodine (I, a and d), gadolinium (Gd, b and e), and Bismuth (Bi, c and f) images generated from the body phantom scans using a single-source (SS)-PCD chess mode acquisition (a–c) and dual-source(DS)-PCD macro-mode acquisition (d–f). The CT image (energy threshold-low image) of the phantom setup is also shown. There are a total of eight vials within the phantom: two containing I (5, 10 mg/mL), two containing Gd (4, 8 mg/mL), two containing Bi (5, 10 mg/mL), as well as two containing mixtures (M1: 5 mg/mL I + 4 mg/mL Gd; M2: 5 mg/mL I + 4 mg/mL Gd + 5 mg/mL Bi). Window/level = 600/100 HU (CT) and 15/7.5 mg/mL (I, Gd, Bi images). [Color figure can be viewed at wileyonlinelibrary.com]

FIG. 5. The root-mean-square-error (RMSE) of mass concentration values for iodine (I), gadolinium (Gd), and bismuth (Bi) in the body phantom images for data acquired using a single-source (SS)-PCD chess mode scan and dual-source (DS)-PCD macro-mode scans. The RMSE calculation included all pixels within the ROIs in different contrast vials. [Color figure can be viewed at wileyonlinelibrary.com]

The improved multi-contrast imaging performance of the DS-PCD approach can potentially benefit scenarios that require simultaneous imaging of multiple nanoparticle-based contrast agents that have distinct K-edges; the successful application of such novel contrasts will rely substantially on accurate material quantification. The DS-PCD approach may also improve the performance of multi-contrast, single phase imaging, which, if successful, may substantially reduce the radiation dose required to obtain clinically acceptable image quality compared to multi-phase scanning.

Several studies have shown that the PCD-CT have various advantages compared to the conventional EID-CT, including improved iodine contrast, reduced electronic noise, more stable CT number in low dose realm, and dose-efficient high-resolution capability.^{6–10} The DS-PCD system is expected to preserve these advantages as it is equipped with the same core PCD technology. These advantages can potentially be used to improve image quality under the same radiation dose, or to reduce patient dose while maintain the same image quality such as resolution and noise level. In addition, the

FIG. 6. The noise power spectra of iodine (I)-, gadolinium (Gd)-, and bismuth (Bi)- specific images generated from data acquired on SS-PCD-CT and DS-PCD-CT. [Color figure can be viewed at wileyonlinelibrary.com]

DS-PCD can be more dose efficient for multi-energy applications as demonstrated here and in a previous work.³⁷

In this work, we used filtered back projection reconstruction to generate the CT images and a least-square based method without denoising to perform material decomposition. This was done to compare the underlying performance of the different approaches without potentially confounding the results by inclusion of other factors, such as iterative reconstruction. The material decomposition performance shown here most likely can be further improved if iterative reconstruction and material decomposition methods that include noise reduction techniques were to be used.^{43–53} Note that DS technology required image-based decomposition methods, which were indeed used for this study.

Similar to the current DSDE-CT systems using EID, a DS-PCD system does not have co-registered projection data between the two subsystems. It is therefore not straightforward how projection domain denoising using co-registered projection data should be performed with DS-PCD. However, image domain spectral denoising techniques can be applied to a DS system to reduce image noise, as have been implemented on commercial CT systems. Such algorithms have been tested in various clinical applications and been shown to

TABLE III. Iodine (I), gadolinium (Gd), and bismuth (Bi) mass concentration measurements (mean $\pm$ standard deviation) for each vial in the body phantom for data acquired using a single-source (SS)-PCD chess mode scan and dual-source (DS)-PCD macro-mode scans.

							Mixture 2 (1.5 mg/mL + Gd 4 mg/mL + Bi 5 mg/mL)	
	I 5 mg/mL	I 10 mg/mL	Gd 4 mg/mL	Gd 8 mg/mL	Bi 5 mg/mL	Bi 10 mg/mL	Gd 4 mg/mL	Bi 5 mg/mL
SS-PCD with double radiation dose (CTDI _{vol,32cm} = 28 mGy)	5.9 $\pm$ 9.1	10.3 $\pm$ 10.7	3.1 $\pm$ 7.2	8.8 $\pm$ 6.2	4.7 $\pm$ 1.3	10.4 $\pm$ 1.7	5.5 $\pm$ 10.0 (I) 3.6 $\pm$ 7.2 (Gd)	7.2 $\pm$ 9.2 (I) 2.6 $\pm$ 6.5 (Gd) 5.3 $\pm$ 1.4 (Bi)
DS-PCD (CTDI _{vol,32cm} = 14 mGy)	5.5 $\pm$ 3.0	10.0 $\pm$ 2.7	3.9 $\pm$ 1.8	8.2 $\pm$ 1.9	5.1 $\pm$ 1.1	10.1 $\pm$ 1.1	5.4 $\pm$ 2.7 (I) 3.6 $\pm$ 2.2 (Gd)	5.5 $\pm$ 2.4 (I) 3.5 $\pm$ 2.0 (Gd) 5.5 $\pm$ 1.2 (Bi)

be effective.^{54–57} The loss of co-registration between DSCT does not necessarily cause loss of spatial resolution, because the two subsystems are synchronized with each other during acquisition, and both of them have the same spatial resolution as a single-source CT.

The performance of multi-contrast material decomposition depends on object size; increased object size (i.e., abdomen vs head-size object) can reduce multi-contrast imaging performance. This is likely due to the hardening of the x-ray beam as the object size increases, which shifts the mean photon energy in each energy bin toward higher energy values and reduces the differences in mean energies between different energy bins. Thus, we evaluated the DS-PCD approach for both head and body phantoms of standard sizes. Imaging patients of different sizes, particularly for very small and very large patients, will require further investigation to optimize the tube potential and energy threshold settings, similar to what has been done for DECT using EID DSCT systems.^{42,58,59}

Similar to studies with EID DECT using DS systems, an additional tin filter was used to further improve energy separation between data acquired using low and high tube potentials.³⁵ However, other filtration materials, such as gold, Gd, or a combination of multiple materials, may also be used to improve the separation of different energy bins.^{60,61} The choice of filtration materials, as well as their thicknesses, must be optimized based on the specific-imaging task and the contrast materials of interest; such optimization might further improve material decomposition performance.

This study was performed based on a research whole-body PCD-CT system. This system reflects the performance of the only whole-body PCD-CT system that is currently available. As a research device, it has limitations that would not be present in a commercial system. For example, the chess mode, with its 50% dose efficiency, was a convenient method of providing simultaneous measurement of four energy thresholds at an early stage in the technology development, when detector pixels with only two energy thresholds were available for use in the project. The chess mode, with its inherent dose penalty, would not be used in a commercial system. With further technical advances in PCD technology, and further optimization of detector and system design based on knowledge obtained with the research system, considerable improvement in PCD-CT performance is anticipated.

One limitation of this study is that cross scattering between two detector arrays when dual-source data are simultaneously acquired was not considered in the phantom experiment. In reality, the cross scattering can reduce the energy separation between the low- and high-energy subsystems, and therefore degrade decomposition performance. On the other hand, cross scattering correction techniques can be considered. On the current dual-source CT systems, cross scattering also exists, and can usually be corrected to an acceptable level.^{62,63} Similar methods could be used for cross scattering correction in the DS-PCD system, although the difference between EID and PCD on the effect of cross scattering is an interesting area that worth further investigation.

The purpose of this preliminary study was to demonstrate the feasibility of performing multi-contrast imaging on a DS PCD platform. To achieve this goal, we scanned contrast solutions with concentrations typically seen in literature.²⁵ This setup allows us to use contrast material RMSE to compare the quantitative performance of DS-PCD and SS-PCD. Detection of low contrast agent concentration may require significant effort on developing appropriate postprocessing pipeline to achieve optimal performance, which may require the use of iterative reconstruction, additional image denoising, and material decomposition technique with noise reduction. Therefore, considerable efforts are required to properly establish the lowest concentration that can be resolved for a particular application of interest.

5. CONCLUSION

In this work, we explored the feasibility of using DS two energy-bin PCD-CT system design to perform simultaneous imaging of I, Gd, and Bi contrast materials, and compared its performance with an experimental whole-body SS-PCD system using chess acquisition mode with four energy bins. Phantom experiment results under an ideal condition without cross scattering demonstrated that DS-PCD reduced material quantification RMSE and reduced noise in the material-specific images compared to the SS-PCD-CT using chess mode, therefore showing the potential of DS-PCD-CT in simultaneous multi-contrast imaging.

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CONFLICT OF INTEREST

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^{a)}Author to whom correspondence should be addressed. Electronic mail: leng.shuai@mayo.edu; Telephone: (507) 284-8500; Fax: (507) 266-3661.

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