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Abstract. Energy-resolved photon-counting-detector CT (PCD-CT) is promising for material decomposition with multiple contrast agents using two or more energy bins. However, corrections for nonidealities of PCDs are required, which are still active research topics. In addition, PCD-CT is also likely to have a very high cost due to the current lack of mass production capabilities. We proposed an alternative approach to perform multienergy CT (MECT), which is achieved by acquiring triple or quadruple x-ray beam measurements on a dual-source CT scanner. This strategy was based on a “twin-beam” design on a single-source scanner for dual-energy CT. Examples of beam filters and spectra for triple and quadruple x-ray beam were provided. Computer simulation studies were performed to evaluate the noise and accuracy of material decomposition for multiple contrast mixtures using both triple- and quadruple-beam configurations, compared with the performance on a PCD-CT platform. The results demonstrated that the image quality and dose efficiency of the triple-beam configuration in the proposed MECT technique were comparable to that in PCD-CT. The proposed technique can be readily implemented on a dual-source scanner, which may allow material decomposition of multiple contrast agents to be performed on clinical CT scanners with energy-integrating detectors. © 2018 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: [10.1117/1.JMI.5.3.033502](https://doi.org/10.1117/1.JMI.5.3.033502)]

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1 Introduction

The ability to image multiple contrast agents simultaneously would benefit many abdominal and cardiovascular applications, such as characterization of the macrophage burden, calcification, and stenosis of atherosclerotic plaques by imaging gold nanoparticles, iodine, and calcifications;^{1,2} improving luminal depiction by differentiating among intravascular gadolinium, calcified plaque, and stent;³ separation of the lumen and bowel wall by imaging iodine and bismuth in CT enterography to improve diagnosis of tumor in bowel wall and luminal filling defects;⁴ and performing multiphase abdominal or vascular imaging in one single scan to reduce radiation dose.^{5–9} These applications have the potential to improve CT diagnosis that may benefit millions of CT patients.

Dual-energy CT (DECT) has gained widespread use for material-specific imaging in many clinical applications over the past decade,^{10–12} due to several successful commercial implementations (e.g., dual-source, fast-kV switching, and dual-layer detector). Using DECT with energy-integrating detectors (EIDs), at most three materials in a mixture can be quantified if an additional physical constraint (e.g., volume or mass conservation) is applied in addition to the dual-energy measurements.^{13,14} Without invoking the additional constraint, DECT can only allow a two-material mixture to be stably resolved. When the number of materials in a mixture is more than two and one or more of the components have distinctive K-edges, it is challenging for DECT to perform accurate and stable material decomposition.^{14,15}

Energy-resolved photon-counting-detector CT (PCD-CT) has received much attention in recent years as a method for

material-specific imaging with multiple contrast agents.^{16–22} However, PCD-CT has limitations, the most significant of which is that PCD-CT detectors may produce severely distorted spectral information due to many nonideal effects,¹⁹ such as pulse pile-up,^{18,23} charge sharing,^{24,25} K-escape,^{24,26} Compton scattering, and charge trapping. As a result, even for traditional dual-energy applications, such as virtual noncontrast imaging and iodine quantification, which only involve two-material mixtures, the performance of material-specific imaging with PCD-CT is similar to or worse than that with DECT with optimized beam separation performed at the same total radiation dose.^{27,28} Correction of these nonideal effects remains a promising research topic¹⁹ but has not been implemented to date on a PCD-CT system capable of clinical imaging. Further, PCD-CT is likely to be associated with a very high cost due to the lack of mass production techniques, which may greatly impede the adoption of PCD-CT in routine practice.

In this work, we will investigate the feasibility of using traditional EID-based CT scanners to perform multienergy CT (MECT) as an alternative to PCD-CT for material-specific imaging of multiple contrast agents. This alternative MECT design is based on a dual-source CT that can acquire three or four unique x-ray spectra nearly simultaneously. In this design, the x-ray beam is prefiltered by two different materials, each of which covers half of the detector rows along the longitudinal direction. This split-filter approach was commercially implemented on a Siemens single-source CT scanner to enable DECT.²⁹ Using a similar design on a dual-source CT can enable MECT. Computer simulations were performed to evaluate the noise and accuracy of material decomposition of multicontrast

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agents for both triple- and quadruple-beam configurations in comparison with a research whole-body PCD-CT system.^{20,21}

2 Methods

2.1 Single-Source CT Geometry with a “Twin-Beam” Design

Figure 1(a) shows a “twin-beam” design proposed by Siemens on a single-source CT scanner to enable DECT capability (Somatom Edge, Siemens Healthcare GmbH, Germany). In this design, the same x-ray beam is prefiltered by two different materials, specifically gold (Au) and tin (Sn), each of which covers half of the detector rows along the longitudinal direction. The x-ray beam filtered by Sn is hardened to form the “high energy,” and the x-ray beam filtered by Au is softened because of the K-edge (80.7 keV) to form the “low energy.” In a helical scan with a low helical pitch (<0.5), dual-energy data for the same anatomic section can be acquired at a time interval of approximately half of the rotation time, which can be as quick as 125 ms if the rotation time is 0.25 s.

Figure 1(b) shows the spectra of the low- and high-energy x-ray beams. The mean energies of the two spectra were 67.5 keV after a 0.05-mm Au filter and 85.3 keV after a 0.6-mm Sn filter. The two spectra after the split filter still have significant overlap, mainly due to limits on filtration from the same 120-kV beam. A higher kV beam (e.g., 140 or 150 kV) would not improve the separation, but instead would result in mean energies

that are too high at both beams, which is not ideal for dual-energy applications, as shown in some of the recent evaluation studies.^{30,31} This situation can be substantially improved using triple- and quadruple-beam configurations proposed below.

2.2 Dual-Source CT Geometry with Triple- or Quadruple-Beam Acquisition

The “twin-beam” configuration with a split-filter design can be readily extended to a dual-source CT scanner by operating one (triple-beam configuration) or both (quadruple-beam configuration) sources in the “twin-beam” mode to acquire three or four distinct x-ray beam measurements.

Figure 2(a) shows a triple-beam configuration, in which one of the sources is operated at the “twin-beam” mode, while the other source is operated in a single-energy mode. Together, three distinct x-ray beam measurements can be obtained. Figure 2(b) shows a quadruple-beam configuration, in which both sources are operated at the “twin-beam” mode. Together, four distinct x-ray beam measurements can be obtained.

2.3 X-Ray Beam and Split Filter in the Triple- or Quadruple-Beam Configuration

The success of MECT using the triple- or quadruple-beam configuration in the proposed design is highly dependent on how well the beam spectra can be separated, which primarily determines the noise properties in the material decomposition.^{13,32}

Fig. 1 (a) Twin-beam configuration on a single-source CT and (b) x-ray spectrum of the twin-beam configuration after beam filtration of Au and Sn.

Fig. 2 (a) Triple-beam configuration on a dual-source CT with a split filter added to one of the sources and (b) quadruple-beam configuration with split filters added to both sources.

Later, we describe the tube potentials of the two x-ray tubes and the filters used in the proposed triple- and quadruple-beam configurations.

For the triple-beam configuration, one of the sources was set at 80 kV and the other one was at 150 kV but filtered by 0.6 mm of Sn and 0.15 mm of Au, respectively. The 0.6 mm of Sn is currently used in the third-generation dual-source CT scanner for the high-kV x-ray beam, which provides a reasonable balance between radiation output and spectral separation. A thicker Sn filtration would result in even harder beam (and better separation from the low-energy beam) but insufficient radiation output capacity for typical patient attenuation. The use of gold as the other filter material was due to the high K-edge of gold at 80.7 keV, which selectively absorbs more high-energy photons and “softens” the beam. The selection of 0.15-mm filter thickness was determined by approximately matching the total energy of the photons to that in the Sn-filtered beam for given tube current.

For the quadruple-beam configuration, split filters were added on both x-ray sources, one at 90 kV and the other at 150 kV. The split filters on the 150 kV were exactly the same as in the triple-beam configuration. The split filter on the 90 kV consists of 0.1-mm Sn and 0.11-mm Gd_2S_2O . The Sn filter will “harden” the beam and the Gd_2S_2O filter will “soften” the beam by utilizing the K-edge of Gd, which is at 50.2 keV.

2.4 Simulation of Triple- and Quadruple-Beam Acquisitions

A simulation tool (DRASIM, Siemens) was used for generating the spectra and the projections for triple- and quadruple-beam configurations. This tool was used previously and has proved to be accurate.^{19,33,34} Note that cross-scattering effect from the two sources and between split-filtered beams was not included.

A cylindrical water phantom in a diameter of 20 cm containing nine inserts, including three iodine solutions (4, 6, and 8 mg/cc of iodine); three bismuth solutions (2, 3, and 4 mg/cc of bismuth); and three iodine/bismuth mixtures (8/4, 6/3, and 4/2 mg/cc of I/Bi), was used in the simulation (Fig. 3). The use of iodine and bismuth contrast agents was to represent the clinical application of MECT in CT enterography to improve diagnosis of tumor in bowel wall and luminal filling defects.⁴

The triple-beam configuration described above was used to generate the three beam measurements, which were subsequently corrected for beam hardening using a calibration water phantom at each corresponding spectrum. The mAs was 160 for the two 150-kV beams and 163 for the 80 kV such that the total tube radiation output was matched among the three beams. The total CTDIvol was 9.8 mGy. Images were reconstructed using a filtered-backprojection method. Detailed information about the tube potential, split filter, and mAs used in the simulation are specified in Table 1, and scanner geometry is specified in Table 2. The total CTDIvol was chosen to be approximately double the typical dose in abdominal CT for a 20-cm patient size to improve the precision of comparison among MECT techniques.

The quadruple-beam configuration was used to generate the four beam measurements. The mAs was 270 for the two 90-kV beams and 120 for the two 150-kV beams. The total tube radiation output for each beam was matched, with a total CTDIvol of 9.8 mGy, the same as that in the triple-beam configuration. Again, detailed information about the tube potential, split filter, and mAs are specified in Table 1, and scanner geometry is specified in Table 2.

Fig. 3 A digital phantom used in the simulation, which contains samples with different concentrations iodine and bismuth.

Table 1 Split filters and mAs for the triple- and quadruple-beam configurations used in the computer simulation.

	kV	mAs	CTDIvol (mGy)	Beam no.	Split-filter material	Split-filter thickness (mm)
Triple beam	80 kV	163	3.3	1	NA	NA
	150 kV	160	3.2	2	Au	0.15
			3.2	3	Sn	0.6
Quadruple beam	90 kV	270	2.5	1	Gd_2O_2S	0.11
			2.5	2	Sn	0.1
	150 kV	120	2.4	3	Au	0.15
			2.4	4	Sn	0.6

Table 2 Scanner geometry used in the computer simulation for the triple- and quadruple-beam configurations.

Focal length	595 mm
Source detector distance	1095.6 mm
Detector shape	Cylindrical
Detector material	Gd_2O_2S
Detector thickness	1.02 g/cm ²
Fan angle	50 deg
No. of channels	736
No. of projections	1152
No. of rows	1
Slice width	0.6 mm
Focus width	0.7 mm
Focus length	7.185 mm
Anode angle	7 deg

2.5 Basis Material Decomposition

Basis material decomposition was performed on images after reconstruction. For each pixel, in addition to multienergy measurements, volume conservation was used as a physical constraint, which yields four linear equations for the triple-beam configuration and five linear equations for the quadruple-beam configuration, which are given as

$$\begin{cases} \mu(E_1) = \sum_{m=1}^M \left(\frac{\mu}{\rho} \right)_m(E_1) \cdot \rho_m \\ \mu(E_2) = \sum_{m=1}^M \left(\frac{\mu}{\rho} \right)_m(E_2) \cdot \rho_m \\ \dots \\ \mu(E_N) = \sum_{m=1}^M \left(\frac{\mu}{\rho} \right)_m(E_N) \cdot \rho_m \\ 1 = \sum_{m=1}^M \frac{\rho_m}{\rho_{m0}} \end{cases}, \quad (1)$$

where $\mu(E_i)$, $i = 1, 2, \dots, N$ denote the linear attenuation coefficient at i 'th beam measurement, $\left(\frac{\mu}{\rho} \right)_m(E_i)$ is the mass attenuation coefficient of the m 'th basis material at i 'th beam measurement, $M = 3$ representing the number of basis materials to be decomposed (e.g., iodine, bismuth, and water), $N = 3$ for triple beams and $N = 4$ for quadruple beams, and ρ_m and ρ_{m0} are the mass density of the m 'th basis material in the mixture and in the pure form, respectively. The coefficient matrix was determined using a calibration procedure on a separate phantom containing another group of samples of iodine, bismuth, and water with different mass densities. The coefficient was fit as the slope of data points plotted with linear attenuation coefficient $\mu(E)$ against the know mass densities ρ_m for each material m .

A weighted least-square optimization method was used to solve the material density at each pixel, with the weighting factor being the inverse of the variance of the measurement. The cost function for the optimization method is given by

$$\text{argmin}(\vec{\mu} - A\vec{\rho})^T V^{-1}(\vec{\mu} - A\vec{\rho}),$$

where A represents the coefficient matrix

$$A = \begin{pmatrix} \left(\frac{\mu}{\rho} \right)_1(E_1) & \left(\frac{\mu}{\rho} \right)_2(E_1) & \left(\frac{\mu}{\rho} \right)_3(E_1) \\ \vdots & \ddots & \vdots \\ \left(\frac{\mu}{\rho} \right)_1(E_N) & \left(\frac{\mu}{\rho} \right)_2(E_N) & \left(\frac{\mu}{\rho} \right)_3(E_N) \end{pmatrix},$$

$$V = I_{N \times N} \begin{pmatrix} \sigma_1^2 \\ \vdots \\ \sigma_N^2 \end{pmatrix} \text{ is the variance matrix for CT measurements}$$

from different beams, $\vec{\rho} = \begin{pmatrix} \rho_1 \\ \rho_2 \\ \rho_3 \end{pmatrix}$, and

$$\vec{\mu} = \begin{pmatrix} \mu(E_1) \\ \vdots \\ \mu(E_N) \end{pmatrix}. \text{ To maintain the original noise properties for}$$

the material decomposition, no noise reduction method was applied.

2.6 Comparison with a PCD-CT System Using Computer Simulations

A PCD simulation tool was used to simulate the photon transport process. A similar tool was used in previous publications.²⁰

Table 3 Photon counting CT geometry used in the simulation (not the same as the true system).

Focal length	595 mm
Source detector distance	1095.6 mm
Detector shape	Cylindrical
Detector material	CdTe
Fan angle	50 deg
Slice width	0.6 mm
kV	140
mAs	100
CTDIvol (mGy)	9.8
Energy threshold (keV)	25, 50, 65, 85

Major nonideal effects were included in the simulation, including pulse pile-up, charge sharing, and K-escape. The detector was based on cadmium telluride (CdTe), and each photon-counting subpixel is $0.225 \text{ mm} \times 0.225 \text{ mm}$. Subpixels are grouped to form a 4×4 macropixel of $0.9 \text{ mm} \times 0.9 \text{ mm}$. The real PCD-CT scanner uses a “chess mode” for each macropixel to generate four energy thresholds, with half of the 16 subpixels operating at 2 energy thresholds and the other half at 2 different thresholds.²¹ The chess mode reduces dose efficiency because each eight subpixel only uses half of the x-ray photons incident on a macropixel. In our simulation, we did not simulate the chess mode since it sacrifices dose efficiency. Instead, our simulation assumed that each subpixel can directly generate 4 energy thresholds, and all 16 subpixels in a macropixel were operating at the same 4 energy thresholds so that no radiation dose was wasted. The four energy thresholds were determined to be 25, 50, 65, and 85 keV. After the four energy thresholds were generated, projection data for four energy bins, [25 50], [50 65], [65 85], and [85 140] keV, were created by subtracting the corresponding threshold data with the energy bin [85 140] equal to the fourth energy threshold. The simulated projection data at each energy bin underwent separate beam-hardening correction before they were reconstructed with the filtered-backprojection method. The geometry of the PCD-CT scanner used in the simulation is specified in Table 3. The digital phantoms and multiple contrast samples described above were used in the comparison. The noise and root-mean-square error (RMSE) of the density maps for each material were compared at equivalent radiation doses.

3 Results

3.1 Spectra of the Triple- and Quadruple-Beam Configurations

Figures 4(a) and 4(b) provide the x-ray beam spectra generated by triple- and quadruple-beam configurations after applying the prefilters. For the triple-beam configuration, one of the sources was set at 80 kV and the other one was at 150 kV but filtered by 0.6 mm of Sn and 0.15 mm of Au, respectively. Because one of the sources was operated separately at 80 kV, the mean energy of the low-kV beam was as low as 52.2 keV, whereas the split filter

Fig. 4 X-ray beam spectra generated by (a) a triple-beam configuration, (b) a quadruple-beam configuration, and (c) a PCD-CT scanner.

applied to the 150-kV beam yielded two relatively higher mean energies: 79.7 and 98.6 keV. The mean energies from this triple-beam configuration cover a reasonable range of diagnostic x-ray beam energies. In the quadruple-beam configuration given in Fig. 4(b), split filters were added on both sources, one at 90 kV and the other at 150 kV, yielding four beams with mean energies of 56.1, 60.8, 79.7, and 98.6 keV.

3.2 Material Decomposition Images for Triple- and Quadruple-Beam Configurations

Figures 5(a)–5(c) show the images reconstructed from the triple-beam acquisitions: 80 kV, 150 kV + Au filter (150 Au), and 150 kV + Sn filter (150 Sn), respectively. As expected, the CT number of the iodine decreased as the x-ray beam became

Fig. 5 Images acquired from the triple-beam configuration: (a) 80 kV, (b) 150 kV + Au filter, (c) 150 kV + Sn filter, and mass density maps of the three materials: (d) iodine, (e) bismuth, and (f) water.

Fig. 6 Images acquired from the quadruple-beam configuration: (a) 90 kV + Gd filter, (b) 90 kV + Sn filter, (c) 150 kV + Au filter, (d) 150 kV + Sn filter, and mass density map of the three materials: (e) iodine, (f) bismuth, and (g) water.

harder. However, the CT number of bismuth was higher at a harder beam at 150 Sn than at 150 Au because of the K-edge. Figures 5(d)–5(f) show the mass density map of the three materials: iodine, bismuth, and water, respectively, after the material decomposition.

Figures 6(a)–6(d) show the images reconstructed from the quadruple-beam acquisitions: 90 kV + Gd filter (90 Gd),

90 kV + Sn filter (90 Sn), 150 kV + Au filter (150 Au), and 150 kV + Sn filter (150 Sn), respectively. Figures 6(e)–6(g) show the mass density maps of the three materials: iodine, bismuth, and water, respectively, after material decomposition. Figures 7(a) and 7(b) plot the mass density of iodine and bismuth for all the samples measured with the triple- and quadruple-beam configurations, respectively, compared with the true concentrations of iodine and bismuth. The error bars indicate the standard deviation of each measurement. Obvious positive bias was observed at zero concentration of either iodine or bismuth. The zero concentration of iodine corresponds to the bismuth samples with a concentration of 2, 3, and 4 mg/cc, whereas the zero concentration of bismuth corresponds to the iodine samples with a concentration of 4, 6, and 8 mg/cc. For nonzero concentrations, the bias was quite small and could be either positive or negative.

3.3 Comparison with PCD-CT

The spectra of the four energy bins from the PCD-CT scanner ([25 50] keV, [50 65] keV, [65 85] keV, and [85 140] keV) are shown in Fig. 4(c). The mean energies of these four energy bins are 63.0, 65.1, 80.4, and 105.3 keV, respectively. One can see that, because of the nonideal spectral separation, significant amount of overlap exist among these energy bins. In particular, the low-energy bin at [25 50] keV was contaminated by low-energy photons originated from interactions between higher-energy photons and PCDs, making the mean energy well beyond the energy window. The spectral separation in the triple- and quadruple-beam configurations appears to be similar or even better (Table 4).

Figures 8(a)–8(d) show the images reconstructed from the four energy bins acquired on the PCD-CT scanner: [25 50], [50 65], [65 85], and [85 140] keV, respectively. Figures 8(e)–8(g) show the mass density maps of the three materials: iodine, bismuth, and water, respectively, after material decomposition. Figure 9 plots the mass density of iodine and bismuth for all the samples measured on the PCD-CT scanner compared with the true concentrations of iodine and bismuth.

Fig. 7 Mass densities of iodine and bismuth samples measured from (a) triple-beam and (b) quadruple-beam acquisitions, compared to the true concentrations of iodine and bismuth. The multiple caps on the error bars were because of the two measurement samples for each concentration value.

Table 4 Mean energies of the spectra in triple- and quadruple-beam configurations compared to PCD-CT.

	Beam 1		Beam 2		Beam 3		Beam 4	
	kV/filter or energy bin	Mean energy	kV/filter or energy bin	Mean energy	kV/filter or energy bin	Mean energy	kV/filter or energy bin	Mean energy
EID-CT triple beam	80 kV	52.2 keV	150 kV + Au	79.7 keV	150 kV + Sn	98.6 keV	NA	NA
EID-CT quadruple beam	90 kV + Gd	56.1 keV	90 kV + Sn	60.8 keV	150 kV + Au	79.7 keV	150 kV + Sn	98.6 keV
PCD-CT	[25, 50] keV	63.0 keV	[50, 65] keV	65.1 keV	[65, 85] keV	80.4 keV	[85, 140] keV	105.3 keV

Fig. 8 Images from the four energy bins on the PCD-CT scanner: (a) [25 50] keV, (b) [50 65] keV, (c) [65 85] keV, and (d) [85 140] keV, and mass density maps of the three materials: (e) iodine, (f) bismuth, and (g) water obtained from material decomposition.

Fig. 9 Mass densities of iodine and bismuth samples measured from the PCD-CT scanner compared to the true concentrations. The multiple caps on the error bars were because of the two measurement samples for each concentration value.

The noise level of the original CT images at each spectrum or energy bin, the noise level of mass density for each basis material after material decomposition, the RMSE of the mass density of each basis material for the triple- and quadruple-beam configurations, and the PCD-CT are provided in Table 5. It can be seen that the noise performance of the triple-beam

configuration is comparable to that of the PCD-CT. The noise from the quadruple beam is slightly higher than the other two configurations, which suggests that adding more energy beams may not necessarily improve the performance.

4 Discussion

MECT has a great potential to image multiple contrast agents simultaneously, which would benefit many abdominal and cardiovascular applications. Currently, some of the major CT manufacturers are devoting their efforts to PCD technology partly based on the belief that PCD-CT is the best approach to MECT. There have been no sufficient studies showing that PCD-CT is better than EID-based MECT for material-specific imaging, and some of the simulation studies actually showed that EID-CT is similar or superior to PCD-CT for DECT applications.^{23,28}

In this work, an alternative technique was proposed to implement the MECT technology using split filters on existing dual-source CT scanner. Similar spectral performance and dose efficiency compared to a PCD-CT system were demonstrated in a computer simulation study, which confirmed the findings in previous studies.²⁸

The proposed EID-based MECT system is expected to have many advantages over a PCD-CT system. First, because the needed hardware changes are minimal (added split filters),

Table 5 Comparison of noise levels in the original images, noise levels in the mass density images, and the RMSE of the mass density measurements among triple-beam, quadruple-beam, and PCD-CT.

	Noise level in CT images (HU)				Noise level in material density maps (mg/cc)			RMSE in material density maps (mg/cc)		
	Beam 1	Beam 2	Beam 3	Beam 4	Iodine	Bismuth	Water	Iodine	Bismuth	Water
EID-CT 3 beams	12.1	11.9	11.9	NA	0.60	0.62	0.074	0.69	0.71	0.076
EID-CT 4 beams	13.6	12.6	13.4	13.8	0.63	0.71	0.074	0.74	0.82	0.075
PCD-CT 4 energy bins	14.5	12.2	12.1	13.6	0.55	0.65	0.062	0.65	0.76	0.065

the implementation will be rather straightforward. Second, compared to a PCD-CT system, the cost of implementation is almost negligible. Third, reasonable spectrum separation can be achieved without the need to worry about some of the nonideal factors, such as charge sharing, k-escape, and pulse pile-up, which have been challenging to overcome in a PCD-CT system. Finally, the dose can be readily distributed among different beams, e.g., high tube current on low tube potentials, whereas the dose distribution among different energy bins is challenging in a PCD-CT system, which is another factor degrading the imaging performance.

The split-filter concept itself is not new since Siemens already implemented it on their single-source CT scanner to enable DECT scanning. Because of the significant spectral overlap, the spectral performance is not ideal, as shown in some recent studies.^{30,31} The use of split filter to enable MECT technology is significant, in which it represents the first scanner platform to implement EID-based MECT on a clinical scanner. Because of its simplicity, the system could readily become a testing platform for material-specific imaging of multiple contrast agents.

The concept of implementing MECT on traditional EID-based CT scanners is not necessarily limited to the dual-source scanner platform. For example, for dual-energy scanner platforms, such as fast-kV switching, it is also possible to implement MECT by switching among 3 or 4 energy spectra. As suggested in this study, it is also possible to achieve similar spectral performance compared to a PCD-CT system.

A potential challenge in the practical implementation of the proposed system is cross scattering from the two sources and between split-filtered beams, which is not included in the simulation software, as stated above. The manufacturer has implemented correction methods using additional detector measurements, which works quite well for DECT and PCD-CT. Therefore, it is not expected to be a major issue. Another limitation is that the proposed MECT system can only be implemented during a helical scan, as shown in Fig. 2. In the current simulation study, we only used a 2-D geometry with one detector row, as specified in Table 2. The reason why we chose to use a 2-D geometry is mainly for simplicity, but we do not anticipate fundamental differences moving from 2-D to 3-D in this technique since the material decomposition method is image based. As long as the CT number is accurate in image reconstruction, the input to the material decomposition will be the same. For proof-of-concept simulation study, we believe simulating in 2-D geometry should be sufficient to demonstrate the noise and spectral performance of the proposed technique. During a helical scan, there will be a mismatch

between the projection data acquired by the beams after the split filter and the two sources, which is the reason why the material decomposition is performed in image domain rather than projection domain. However, some recent development of iterative reconstruction has the potential to enable material decomposition in projection space for this data acquisition geometry.³⁵

With the prior information (e.g., volume conservation) incorporated into the material decomposition process, dual-energy data or PCD data with two energy bins are also capable of performing three-material decomposition, such as iodine, bismuth, and water in this study. A systematic comparison among the existing dual-source DECT technique, the proposed EID-based MECT technique, and the PCD-based MECT with two and four energy thresholds, with and without volume conservation, in both computer simulation and experimental studies is currently being investigated.

Practical implementation of the proposed system is underway.

5 Conclusions

We proposed an alternative scanner design to perform MECT, which was achieved by acquiring triple or quadruple x-ray beam measurements on a dual-source CT scanner. Simulation studies showed that the image quality and dose efficiency of the triple-beam configuration were comparable to a PCD-CT scanner for an imaging task involving multiple contrast agents. This design can be readily implemented on a dual-source scanner, which may allow material decomposition of multicontrast agents to be performed on CT scanners with EID.

Disclosures

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