Improved dual-energy material discrimination for dual-source CT by means of additional spectral filtration

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The use of additional spectral filtration for dual-energy (DE) imaging using a dual-source CT (DSCT) system was investigated and its effect on the material-specific DE_{ratio} was evaluated for several clinically relevant materials. The x-ray spectra, data acquisition, and reconstruction processes for a DSCT system (Siemens Definition) were simulated using information provided by the system manufacturer, resulting in virtual DE images. The factory-installed filtration for the 80 kV spectrum was left unchanged to avoid any further reductions in tube output, and only the filtration for the high-energy spectrum was modified. Only practical single-element filter materials within the atomic number range of $40 \leq Z \leq 83$ were evaluated, with the aim of maximizing the separation between the two spectra, while maintaining similar noise levels for high- and low-energy images acquired at the same tube current. The differences between mean energies and the ratio of the 140 and 80 kV detector signals, each integrated below 80 keV, were evaluated. The simulations were performed for three attenuation scenarios: Head, body, and large body. The large body scenario was evaluated for the DE acquisition mode using the 100 and 140 kV spectra. The DE_{ratio} for calcium hydroxyapatite (simulating bone or calcifications), iodine, and iron were determined for CT images simulated using the modified and factory-installed filtration. Several filter materials were found to perform well at proper thicknesses, with tin being a good practical choice. When image noise was matched between the low- and high-energy images, the spectral difference in mean absorbed energy using tin was increased from 25.7 to 42.7 keV (head), from 28.6 to 44.1 keV (body), and from 20.2 to 30.2 keV (large body). The overlap of the signal spectra for energies below 80 keV was reduced from 78% to 31% (head), from 93% to 27% (body), and from 106% to 79% (large body). The DE_{ratio} for the body attenuation scenario increased from 1.45 to 1.91 (calcium), from 1.84 to 3.39 (iodine), and from 1.73 to 2.93 (iron) with the additional tin filtration compared to the factory filtration. This use of additional filtration for one of the x-ray tubes used in dual-source DECT dramatically increased the difference between material-specific DE ratios, e.g., from 0.39 to 1.48 for calcium and iodine or from 0.28 to 1.02 for calcium and iron. Because the ability to discriminate between different materials in DE imaging depends primarily on the differences in DE ratios, this increase is expected to improve the performance of any material-specific DECT imaging task. Furthermore, for the large patient size and in conjunction with a 100/140 kV acquisition, the use of additional filtration decreased noise in the low-energy images and increased contrast in the DE image relative to that obtained with 80/140 kV and no additional filtration. © 2009 American Association of Physicists in Medicine. [DOI: [10.1118/1.3083567](https://doi.org/10.1118/1.3083567)]

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I. INTRODUCTION

Dual-energy (DE) imaging was introduced more than 30 years ago and was applied both to dual-energy CT (DECT) and dual-energy radiography (DER). The theoretical basis for DE material decomposition was described in early works by Alvarez and Macovski¹ and Kalender *et al.*² for DECT, and Lehmann *et al.*³ for DER. While clinical interest in DER has been maintained over the years, especially for chest imaging and bone densitometry, DECT was not widely adopted due to the technical limitations of earlier CT systems. Recently, DECT imaging has been reintroduced into clinical practice.^{4–13} Different approaches for DECT imaging include dual-source CT,⁵ rapid kV switching,^{2,14} and dual-layer

(sandwich) detectors.^{4,15} The dual-source CT approach became commercially available in 2006.^{5,6}

The dual-source CT (DSCT) scanner (SOMATOM Definition, Siemens Healthcare, Forchheim, Germany) has two x-ray tubes and two corresponding arrays of detectors.⁵ Both tubes have the same filtration and generate identical x-ray spectra when operated at the same tube potential. Identical filtration is necessary for dual-source applications where the data from both tubes are combined either to improve the temporal resolution (cardiac mode) or increase the available x-ray power (obese mode). However, using the same filtration for both tubes provides less than optimal conditions for DE acquisitions, where the two tubes are operated at

different tube potentials. The x-ray spectra generated at low (80 kV) and high (140 kV) peak tube potentials have a high degree of spectral overlap, resulting in a separation between the average energies of the two spectra much less than the 60 keV difference in peak potential.

The ability of DECT to discriminate between two materials relies primarily on the difference between the DE_{ratio} of the materials (see Sec. II A), which is determined by the separation between the high- and low-energy spectra and the difference between the effective atomic numbers of the evaluated materials. The smaller the spectral separation, the harder it is to discriminate between two materials, especially for materials with close atomic numbers (e.g., calcium and iron). In the limit of no separation between the two spectra (single-energy limit), no material discrimination is possible if the density differences adequately offset the differences in attenuation due to atomic number.

Spectral separation can be increased by using additional filtration for one or both tube potentials (kV). Previous research on optimizing the added filtration for DE imaging focused on DER applications, primarily on chest radiography¹⁶ and mammography.¹⁷ In these studies, added filtration was optimized for a very specific application, such as detectability of coronary artery calcium¹⁸ or lung nodules.¹⁹ By defining the specific imaging task, researchers were able to select the most relevant performance metric (figure of merit) dependent on several parameters (added filtration, image quality, dose allocation, kV settings, and tube loading) and then find an optimal combination of parameters to maximize the figure of merit.

In contrast to DER, few studies have been published regarding optimization of the added filtration for DECT. In 1979, Kelcz *et al.*²⁰ was the first to emphasize the importance of additional filtration for DECT performed using two unique tube potentials. Two additional DECT studies examined the use of a split filter using a single source scanner²¹ and a prereconstruction technique that changed beam filtration manually every 8 s.²²

The purpose of this study was to determine how additional filtration could improve DE imaging performed using DSCT for the purpose of material discrimination. To achieve this, we evaluated the effect of added filtration on the DE_{ratio} of several clinically relevant materials. In contrast to DER studies, the added filtration was not optimized for a specific DECT imaging task because all DECT applications, regardless of the imaging task, benefit from improved separation between the high- and low-energy spectra. As shown in Sec. II A, the evaluated performance metric, the DE_{ratio} , is directly related to noise in material-specific DE images and, hence, the required radiation dose.

II. MATERIALS AND METHODS

II.A. DE ratio and noise in material-specific images

In DECT, the objectives are (1) to differentiate two materials and (2) to quantify the mass density of each material (herein we use ρ to represent mass density, defined as the mass of a given material per fixed volume, i.e., per voxel).

Assuming that both materials are in a common background material, the energy-dependent linear attenuation coefficient of each voxel $\mu(E)$ can be expressed as the linear combination of

$$\mu(E) = \left(\frac{\mu}{\rho}\right)_1(E) \cdot \rho_1 + \left(\frac{\mu}{\rho}\right)_2(E) \cdot \rho_2 + \left(\frac{\mu}{\rho}\right)_w(E) \cdot \rho_w, \quad (1)$$

where $(\mu/\rho)_i(E)$ and ρ_i ($i=1,2,w$) denote the mass attenuation coefficients and mass densities of the two materials ($i=1,2$) to be differentiated and the background material, here assumed to be water ($i=w$). For simplicity, we consider here only the monoenergetic case, although the same principles apply to the polyenergetic case.

Following the approach used by Kelcz *et al.*,²⁰ the background material density is a function of the other two densities due to the displacement effect and can be approximated as

$$\rho_w(\rho_1, \rho_2) = \rho_{w0} + \rho_1 \frac{\partial \rho_w(\rho_1, \rho_2)}{\partial \rho_1} + \rho_2 \frac{\partial \rho_w(\rho_1, \rho_2)}{\partial \rho_2}, \quad (2)$$

where ρ_{w0} denotes the mass density of the background material when no other materials are present and $\partial \rho_w(\rho_1, \rho_2)/\partial \rho_1$ and $\partial \rho_w(\rho_1, \rho_2)/\partial \rho_2$ are the partial derivatives of the background material density with respect to the densities of the other two materials. Therefore, the CT numbers of the mixture at low- and high-energies are given by

$$CT(E_{low}) = \frac{1000}{\mu_w(E_{low})} [M_1(E_{low}) \cdot \rho_1 + M_2(E_{low}) \cdot \rho_2], \quad (3a)$$

$$CT(E_{high}) = \frac{1000}{\mu_w(E_{high})} [M_1(E_{high}) \cdot \rho_1 + M_2(E_{high}) \cdot \rho_2], \quad (3b)$$

where

$$M_i(E_j) = \left(\frac{\mu}{\rho}\right)_i(E_j) + \frac{\partial \rho_w(\rho_1, \rho_2)}{\partial \rho_i},$$

$i=1,2$, representing the two materials and $j=low, high$ representing the two energies. Defining $\Gamma(E_j) = \mu_w(E_j) \cdot CT(E_j)/1000$, the linear equations (3) can be solved to obtain the mass densities of the two materials

$$\rho_1 = \frac{\Gamma(E_{low}) - \Gamma(E_{high}) \cdot R_2}{M_1(E_{high}) \cdot (R_1 - R_2)}, \quad (4a)$$

$$\rho_2 = \frac{\Gamma(E_{high}) \cdot R_1 - \Gamma(E_{low})}{M_2(E_{high}) \cdot (R_1 - R_2)}, \quad (4b)$$

where $R_1 = M_1(E_{low})/M_1(E_{high})$ and $R_2 = M_2(E_{low})/M_2(E_{high})$. Since the displacement effect is typically quite small, $\partial \rho_w(\rho_1, \rho_2)/\partial \rho_i$ is essentially constant. Hence, R_1 and R_2 , which are proportional to the DE_{ratio} , represent density-independent material-specific parameters. Experimentally, the DE_{ratio} is obtained by measuring the low- and high-energy CT numbers for several different density values of a given material and calculating the ratio of the slopes for the

$CT(E_{low,high})$ versus density lines. The DE_{ratio} measured in this way is proportional to R in

$$R = (\mu_w(E_{low})/\mu_w(E_{high}))DE_{ratio}. \quad (5)$$

Noise present in the original low and high kV images propagates into the material-specific parameters (e.g., mass density), affecting the precision of the results. If we denote the noise in $\Gamma(E_{low})$ and $\Gamma(E_{high})$ as σ_{low} and σ_{high} , respectively, the noise in the resultant mass density images is

$$\sigma_1 = \frac{\mu_w(E_{high})}{\mu_w(E_{low})} \frac{\sqrt{\sigma_{low}^2 + \sigma_{high}^2 \cdot R_2^2}}{M_1(E_{high}) \cdot (DE_{ratio1} - DE_{ratio2})}, \quad (6a)$$

$$\sigma_2 = \frac{\mu_w(E_{high})}{\mu_w(E_{low})} \frac{\sqrt{\sigma_{high}^2 \cdot R_1^2 + \sigma_{low}^2}}{M_2(E_{high}) \cdot (DE_{ratio1} - DE_{ratio2})}. \quad (6b)$$

This result is similar to that obtained by Kelcz *et al.*²⁰ in 1979. It demonstrates that the noise in the calculated material-specific parameters (e.g., mass density) of either of the two materials is inversely proportional to the difference between the DE ratios of the two materials, $DE_{ratio1} - DE_{ratio2}$. This is a very important result, which directly relates the DE ratios of the two materials with the image quality of the DE processed material-specific images. A small difference between the DE ratios results in a significant noise increase. For the task of differentiating two materials, the increased noise leads to a larger uncertainty in distinguishing the two materials. For the task of quantifying the mass density of the two materials, the precision of the mass density information will be similarly compromised. As a consequence, a higher radiation dose becomes necessary in order to achieve a reasonable contrast-to-noise ratio and, hence, certainty in the material-specific DE images. Therefore, the quality of any DECT material-specific images is improved and the required radiation dose decreased, as the difference between DE ratios of the materials of interest is increased.

II.B. Spectral simulation

X-ray tube spectra, which consist of bremsstrahlung and characteristic radiation from a tungsten alloy target, can be accurately simulated using existing theoretical models,²³ or estimated by interpolating polynomials over experimentally measured data.²⁴ We used a previously validated, software simulation tool (DRASIM, Version 8.0, Siemens Healthcare)²⁵ to simulate x-ray transmission of the modeled spectra through virtual phantoms of arbitrary shape and composition. For various tube potentials, the software simulated the spectral attenuation by various absorber materials, taking into account the attenuation path length, shape, and elemental composition of the materials. For all rays between the x-ray focal spot and the detector, DRASIM calculated the attenuation in the beam filter, the phantom, and the antiscatter grids. The x-ray focal spot size and shape, detector bin size, and aperture were also taken into account. From this, the intensities of transmitted radiation were calculated as a function of energy. Knowing the transmitted spectrum and the detector response function, the detector signal (absorbed energy re-

corded by the detector) was determined. This allowed prediction of the spectra at each stage of the imaging chain.

Because the signal produced by a CT detector (*Signal*) is proportional to the energy absorbed in the detector, all optimizations were performed using the absorbed energy spectra, hence,

$$Signal_i(E) \propto S_i(E)D(E)E, \quad (7)$$

where i is the tube potential (e.g., 80, 140), $S_i(E)$ are the x-ray tube spectra (photon number per keV) at the detector, and $D(E)$ is the fractional energy absorption of the detector. The resultant signal spectra represent the energy absorbed by the detector as a function of photon energy. Pragmatically, the actual detector signal is the integration of the signal spectra over all energies.

II.C. Image simulation

Using the x-ray spectra predicted by DRASIM, one can simulate the entire set of CT projection data for any object with known x-ray attenuation properties (i.e., a virtual phantom), reconstruct the projection data to obtain a CT image of the virtual phantom, and measure image noise and the CT numbers of materials of interest. Using this approach and CT simulation software (CT_SIM, Version 9.5, Siemens Healthcare), we produced virtual DE images for the modified and factory-installed filtration. The images were reconstructed using a conventional fan-beam filtered-backprojection algorithm and Hanning kernel. Beam-hardening correction was performed to correct the line integral before image reconstruction.

II.D. Candidate filter materials

Additional filtration of the low-energy spectrum would further decrease tube output, which is already insufficient (even at maximum tube current) for very large patients. For this reason, we left the low-energy spectrum unchanged and used additional filtration only for the high-energy tube, as previously described by Kelcz *et al.*,²⁰ in order to obtain better spectral separation. For additional filtration of the high-energy beam, we considered only practical (nonradioactive, nonvolatile, solid at room temperature, machinable, etc.) single-element materials within the atomic number range of $40 \leq Z \leq 83$. Fifteen filter materials meeting these criteria were evaluated with the aim of maximizing the separation between the two spectra.

II.E. Virtual phantoms

The evaluations were performed using three virtual phantoms representing a patient head, body, and large body. The head phantom was simulated by a 15 cm diameter water cylinder with a 5 mm bone rim. The body phantoms emulated a standard and large size anthropomorphic thorax phantom (QRM, Möhrendorf, Germany). The standard thorax phantom had a lateral dimension of 30 cm, while the large phantom had a lateral width of 40 cm. The 10 cm central cardiac region was mimicked by water. For the head and

TABLE I. Potential candidates for filter materials and the corresponding results under the condition of matched detector signal.

Filter element	Attenuation and DE tube potentials								
	Filter thickness (mm)			Separation (keV)			Overlap (%)		
	Head 80/140 kV	Body 80/140 kV	Large body 100/140 kV	Head 80/140 kV	Body 80/140 kV	Large body 100/140 kV	Head 80/140 kV	Body 80/140 kV	Large body 100/140 kV
Titanium (Ti) ^a	0.9	0.9	0.9	25.7	28.6	20.2	43.5	34.7	56.4
Zirconium (Zr) ^b	2.9	3.9	2.1	49.8	51.9	35.7	3.0	0.9	26.9
Niobium (Nb) ^b	2.1	2.7	1.5	50.1	51.7	35.8	2.7	1.0	26.3
Molybdenum (Mo) ^b	1.6	2.2	1.2	49.8	52.1	35.9	3.2	0.9	25.6
Silver (Ag) ^b	1.2	1.6	0.9	50.2	52.2	36.3	2.7	0.8	24.0
Cadmium (Cd) ^b	1.4	1.9	1.0	50.2	52.3	35.9	2.8	0.8	26.7
Indium (In) ^b	1.6	2.1	1.2	50.4	52.2	36.5	2.5	0.8	23.3
Tin (Sn) ^b	1.5	2.0	1.1	50.2	52.2	36.1	2.8	0.9	25.5

^aCurrent factory-supplied filtration, which also includes 3.0 mm of aluminum. The mAs at 140 kV was reduced to match the detector signal at 80 kV. The mAs_{80}/mAs_{140} ratios were 9.8, 13.2, and 4.6 for the head, body, and large body phantoms, respectively.

^bAdditional filtration for the 140 kV tube only. The same mAs value on both tubes.

body attenuations, the spectral optimization was performed for 80/140 kV DE acquisitions. For the large body scenario, spectral separation was optimized for 100/140 kV acquisitions in order to allow increased x-ray output from the low-energy tube.

II.F. Determination of optimal filter thickness

For every evaluated filter material, the thickness was varied in 0.1 mm increments, and simulated spectra and DECT images were produced for all three virtual phantoms with the aim of matching image noise in the low- and high-energy images. For each virtual phantom, two optimal filter thicknesses were determined corresponding to two noise matching strategies.

In the first scenario, we assumed that by matching the detector signal from the low- and high-energy beams, the noise in the low- and high-energy images would be matched. Thus, we iteratively adjusted filter thickness until the detector signal [area under the signal spectra curve, Eq. (7)] was matched between the high- and low-energy spectra. In practice, however, this assumption is not valid because matching the detector signal does not provide the same statistical noise. This occurs because the signal is directly proportional to photon energy, as shown in Eq. (7). As a consequence, fewer high-energy photons are required compared to low-energy photons to achieve the same signal, resulting in less quantum noise for the low-energy images when detector signal is matched.

In the second scenario, we used the simulated DE image data to evaluate the noise propagated into the high- and low-energy images. Then, we iteratively adjusted filter thickness until noise in the simulated images was matched between the high- and low-energy images. All other parameters (except the filter thickness) were equivalent between the two scenarios. The second scenario was used for making recommendations regarding practical filter thicknesses.

Matching noise between high- and low-energy images was not strictly required. Depending on different dual-energy

processing tasks, such as generating iodine maps, creating linearly mixed images, or synthesizing monochromatic images, the weighting factors for low- and high-energy data are not always the same. This means that the noise levels in low- and high-energy images do not necessarily have to be the same in order to achieve the lowest noise in the DECT images. Since we did not restrict the filter optimization to a specific dual-energy task, matched noise between low- and high-energy images is a reasonable condition for the purpose of investigating the effect of added filtration.

For the current factory filtration, at the maximum output of both tubes, the noise in images formed from a combination of the low- and high-energy images is strongly dominated by the low-energy spectra. As filtration is added to the high-energy tube to improve the spectral separation, image noise would eventually become dominated by the strongly attenuated high-energy spectrum. For the evaluated scanner, the maximum tube currents are approximately equal for the 80 and 140 kV beams. Hence, we imposed the requirement that the proper thickness for the 140 kV beam filtration was the value that produced equivalent image noise in the low- and high-energy images at the same mAs values.

For comparison with the factory filtration, the same two scenarios were examined (matching the detected signal from the low- and high-energy tubes or matching noise between the low- and high-energy images). However, because the filter thickness was fixed and the same on both tubes, the mAs values on each tube were varied in order to match signal or match noise. The appropriate mAs_{80}/mAs_{140} ratios are shown in the footnotes of Tables I and II.

II.G. Determination of signal spectra separation

Using the optimal filter thicknesses and materials for the high-energy tube, the signal spectra were quantitatively assessed using two different metrics, separation ΔE and overlap O . The separation between the average energies of the signal spectra for the 80 and 140 kV spectra was calculated as

TABLE II. Potential candidates for filter materials and the corresponding results under the condition of matched image noise.

Filter element	Attenuation and DE tube potentials								
	Filter thickness (mm)			Separation (keV)			Overlap (%)		
	Head 80/140 kV	Body 80/140 kV	Large body 100/140 kV	Head 80/140 kV	Body 80/140 kV	Large body 100/140 kV	Head 80/140 kV	Body 80/140 kV	Large body 100/140 kV
Titanium (Ti) ^a	0.9	0.9	0.9	25.7	28.6	20.2	78	93	106
Zirconium (Zr) ^b	1.4	1.6	1.0	42.7	44.1	30.2	30	26	78
Niobium (Nb) ^b	1.0	1.1	0.7	42.8	43.9	30.2	29	28	78
Molybdenum (Mo) ^b	0.8	0.9	0.6	43.0	44.2	30.8	28	25	72
Silver (Ag) ^b	0.6	0.6	0.4	43.4	43.6	30.3	26	31	78
Cadmium (Cd) ^b	0.7	0.8	0.5	43.4	44.7	30.7	26	22	73
Indium (In) ^b	0.7	0.8	0.5	42.3	43.8	30.0	34	30	82
Tin (Sn) ^b	0.7	0.8	0.5	42.7	44.1	30.2	31	27	79

^aCurrent factory-supplied filtration, which also includes 3.0 mm of aluminum. The mAs at 140 kV was reduced to match the detector signal at 80 kV. The mAs₈₀/mAs₁₄₀ ratios were 5.4, 4.8, and 2.4 for the head, body, and large body phantoms, respectively.

^bAdditional filtration for the 140 kV tube only. The same mAs value on both tubes.

$$\Delta E = \bar{E}_{140} - \bar{E}_{80} = \frac{\int_0^{140} \text{Signal}_{140}(E) dE}{\int_0^{140} \text{Signal}_{140}(E) dE} - \frac{\int_0^{80} \text{Signal}_{80}(E) dE}{\int_0^{80} \text{Signal}_{80}(E) dE}. \quad (8)$$

The overlap of the signal spectra below 80 keV was obtained using

$$O = \frac{\int_0^{80} \text{Signal}_{140}(E) dE}{\int_0^{80} \text{Signal}_{80}(E) dE}. \quad (9)$$

For the large body attenuation, Eqs. (8) and (9) were modified to correspond to the use of 100 and 140 kV. Because the amount of attenuation (and, hence, the transmitted spectra) varies for different projections through an object, the spectral evaluation was performed for the most attenuating path through each phantom.

II.H. Effect of improved separation on DE ratios

After determining the optimal filter thickness and material (taking into account image noise and practical aspects of the material, such as cost and machinability), we evaluated the effect of the optimized spectra on the DE_{ratio} for three clinically relevant materials: Calcium hydroxyapatite (hereafter referred to as calcium), iodine, and iron. To determine the DE_{ratio}, ten cylinders (10 mm in diameter) were added to the central portion of the virtual thorax phantoms, simulating a combination of five different concentrations of two different materials [Fig. 1(a)]. The two material combinations studied were calcium and iodine, and calcium and iron.

For each of the two combinations of materials, five scan conditions were simulated. The five scan conditions were as follows:

- (1) 30 cm thorax, 80/140 kV, factory-installed filtration;
- (2) 30 cm thorax, 80/140 kV, filtration optimized for 30 cm body attenuation;
- (3) 40 cm thorax, 100/140 kV, factory-installed filtration;

- (4) 40 cm thorax, 100/140 kV, filtration optimized for the 40 cm body attenuation; and
- (5) 30 cm thorax, 80/140 kV, filtration optimized for the 40 cm body attenuation.

Condition 5 was evaluated to determine the trade-off of DE_{ratio} separation improvement versus the technical complexity of having different thickness filters for different patient sizes.

For every simulated DE image, the mean CT number in each cylinder was measured and the CT numbers versus material concentration plotted for both tube potentials. Linear regression was used to determine the DE_{ratio}.

II.I. Experimental validation

For experimental validation of the DE_{ratio} predicted using simulations, the standard and large anthropomorphic thorax phantoms were used in order to match their virtual counterparts, which were used for the simulations. The 10 cm central region contained a water-filled cylinder and a custom Styrofoam frame. The frame was used to hold five 3 cm³ syringes filled with different known concentrations of iodine, and five cylinders with known calcium concentrations (courtesy of QRM). The syringes and the calcium cylinders were approximately 10 mm in diameter, very close to the diameter of the simulated cylinders. The iodine solutions were prepared by diluting iodine contrast medium (Omnipaque 350, GE Healthcare) with water and had iodine concentrations from 3.5 to 17.5 mg/cm³. The concentration of calcium in the QRM cylinders ranged from 50 to 800 mg/cm³. Phantoms were scanned on a dual-source CT scanner using the 80/140 kV acquisition mode (standard phantom) and the 100/140 kV mode (large phantom). One of the reconstructed 5 mm images of the large phantom is shown in Fig. 1(b). The experimental DE_{ratio} for calcium and iodine corresponding to the factory-installed filtration were obtained from the reconstructed DE images using the same procedure as for the

(a)

(b)

FIG. 1. A simulated (a) and experimental (b) high-energy (140 kV) CT image of the large anthropomorphic thorax phantom used in this study. The standard phantom had a lateral dimension of 30 cm and was used for the 80/140 kV simulations. The large thorax phantom had an additional attenuating layer, extending its lateral size to 40 cm; this large phantom was used for the 100/140 kV simulations. The ten cylindrical inserts in the central water-filled portion of the phantom contained five different concentrations for each of two evaluated materials (calcium and iodine shown).

simulated DE images. These results were compared to the respective simulation results in order to validate the accuracy of the simulation techniques.

III. RESULTS

Seven out of fifteen materials were found to perform similarly well at the thicknesses appropriate for matching low- and high-energy image noise for the same tube current. The other eight materials showed significant overlap for the high- and low-energy spectra, most of the time due to the presence of the *K* edge, as shown in Fig. 2. Table I summarizes the results obtained for the first condition, where the detector signal for the high- and low-energy spectra was matched for three different attenuation scenarios: Head (80/140 kV), body (80/140 kV), and large body (100/140 kV). Spectral

FIG. 2. Spectra for the eight materials not chosen as potential candidates for optimal filtration. These spectra were simulated for the virtual body phantom under the condition of matched detector signal. Five of these materials (Pb, Au, Bi, W, and Ta) have a prominent *K* edge. The other three materials (Ce, Gd, and Sm) do not show a prominent *K* edge but still have significantly less separation and more overlap between the high- and low-energy spectra compared to the selected seven “best” materials.

separation was increased from 25.7 to 49.8–50.4 keV (head), from 28.6 to 51.7–52.3 keV (body), and from 20.2 to 35.7–36.5 keV (large body). The spectral overlap was reduced from 43.5% to 2.5%–3.2% (head), from 34.7% to 0.8%–1.0% (body), and from 56.4% to 23%–27% (large body), where the ranges of values correspond to different filter materials. The corresponding DE spectra are shown in Fig. 3.

Table II summarizes the results obtained for the second scenario, where image noise was matched for the low- and high-energy images for three different attenuation scenarios. Spectral separation was increased from 25.7 to 42.3–43.4 keV (head), from 28.6 to 43.6–44.7 keV (body), and from 20.2 to 30.0–30.8 keV (large body). The spectral overlap was reduced from 78% to 26%–34% (head), from 93% to 22%–31% (body), and from 106% to 72%–82% (large body), where the ranges of values correspond to different filter materials. The corresponding DE spectra are shown in Fig. 4.

Since the optimized high-energy spectra for all seven materials were almost identical, DE_{ratio} values were calculated only for tin, which was considered to be the most practical filter material. For the 80/140 kV simulations using the 0.8 mm tin filter and the 30 cm body phantom, the DE_{ratio} increased from 1.45 to 1.91 (calcium), from 1.84 to 3.39 (iodine), and from 1.73 to 2.93 (iron) for the optimized filtration compared to the factory filtration. For the 100/140 kV simulations using the 0.5 mm tin filter and the 40 cm body phantom, the DE_{ratio} increased from 1.23 to 1.48 (calcium), from 1.46 to 2.24 (iodine) and from 1.36 to 1.99 (iron). Finally, for the 80/140 kV simulations using the 0.5 mm tin filter and the 30 cm body phantom, the DE_{ratio} increased from 1.45 to 1.81 (calcium), from 1.84 to 2.96 (iodine), and from 1.73 to 2.67 (iron). The DE_{ratio} results are summarized in Table III. An example of DE_{ratio} calculations is shown in Fig. 5 for the

FIG. 3. Spectra for the first scenario, which required that the detector signal be matched between the low- and high-energy beams. The high- (140 kV) and low- (80 or 100 kV) energy spectra simulated for the current factory-installed filtration [(a)–(c)] and with the additional filtration for the high-energy spectrum [(d)–(f)] using the seven filter candidates (Table 1) are shown. Three different attenuation scenarios are represented: Head [(a) and (d)], body [(b) and (e)], and large body [(c) and (f)] virtual phantoms.

80/140 kV simulations using the 30 cm body phantom containing calcium and iodine inserts. The DE_{ratio} was obtained by calculating the ratios of the corresponding slopes.

The experimental measurements performed using the factory-installed filtration resulted in the DE_{ratio} of 1.54 (cal-

cium) and 1.92 (iodine) for the 30 cm body phantom (80/140 kV) and 1.28 (calcium) and 1.51 (iodine) for the 40 cm body phantom (100/140 kV). The simulated DE_{ratio} had systematically lower values compared to the experimental DE_{ratio} , but the differences were within 6%. The difference between the

FIG. 4. Spectra for the second scenario, which required that the low- and high-energy images had equivalent noise. The high- (140 kV) and low- (80 or 100 kV) energy spectra simulated for the current factory-installed filtration [(a)–(c)] and with the additional filtration for the high-energy spectrum [(d)–(f)] using the seven filter candidates (Table II) are shown. Three different attenuation scenarios are represented: Head [(a) and (d)], body [(b) and (e)], and large body [(c) and (f)] virtual phantoms.

DE_{ratio} for calcium and iodine was essentially the same between the simulations and experiment: 0.39 (simulation) versus 0.38 (experiment) for the 30 cm body phantom, and 0.23 (simulation) versus 0.23 (experiment) for the 40 cm body phantom.

IV. DISCUSSION

Drawing an analogy between the DE_{ratio} and the single-energy CT number, the difference in the DE_{ratio} determines the contrast between two materials in DECT, just as the dif-

TABLE III. DE ratios for three clinically relevant materials (calcium, iodine, and iron) using the factory-installed filtration and added filtration for the high-energy beam. The five data columns correspond to the five scan conditions described in Sec. II H.

Material	Body 80/140 kV			Large body 100/140 kV	
	Factory-installed filtration	Added filtration (tin 0.8 mm)	Added filtration (tin 0.5 mm)	Factory-installed filtration	Added filtration (tin 0.5 mm)
Ca	1.45	1.91	1.81	1.23	1.48
I	1.84	3.39	2.96	1.46	2.24
Fe	1.73	2.93	2.67	1.36	1.99

ference in CT numbers determines the contrast between two materials in single-energy CT. The results of our study demonstrate that the use of appropriate added filtration for the high-energy x-ray tube of the DSCT scanner can dramatically increase the “DE contrast” between two materials, as shown in Table IV. For example, the difference between the DE_{ratio} for calcium and iodine calculated using the 30 cm body phantom and 80/140 kV increased from 0.39 to 1.15 (0.5 mm of tin) or 1.48 (0.8 mm of tin). Such a large increase in DE contrast can significantly enhance the performance of DE algorithms designed to discriminate between calcium (e.g., bone or calcified plaque) and iodinated contrast material. Consequently, this increase in DE contrast can improve the clinical value of DE applications, such as DECT angiography with automatic bone and plaque removal (for better visualization of the vessels).

Improved performance of DE material discrimination suggests possibilities for the development of new advanced clinical applications. One potential application is the noninvasive detection of vulnerable plaques—before they rupture and cause myocardial infarction. Vulnerable plaques are characterized by abnormal proliferation of vasa vasorum (microvessels running inside the vessel wall) and by intraplaque hemorrhage resulting in the accumulation of iron.^{26,27} A major challenge in using iron as an imaging marker of vulnerable plaque is the colocalization of iron and calcium, which are indistinguishable on magnetic resonance imaging and conventional (single-energy) CT, the two most accepted modalities for noninvasive imaging of coronary arteries. However, discriminating hemorrhagic iron from colocalized calcium inside individual plaques might be possible using DECT, provided that the specificity of the technique is sufficient to discriminate these two materials. The results of our study indicate that the difference between the DE_{ratio} for calcium and iron increased from 0.28 to 0.86 (0.5 mm of tin) or 1.02 (0.8 mm of tin) for the 30 cm body phantom (80/140 kV) when the factory-installed filtration was replaced with the optimized filtration. This increase in DE contrast should significantly improve the specificity of a DECT technique for the discrimination of iron versus calcium in vascular plaque.

Finally, the use of optimized filtration can substantially reduce the number of large patients unable to be imaged with DECT due to unacceptable noise in the 80 kV images. These patients potentially could be imaged using 100/140 kV, as this will produce a much larger photon flux for the low-

FIG. 5. Plots of the CT number vs material concentration data for calcium (a) and iodine (b) obtained using simulations (solid lines) and experimental measurements (dashed lines; factory filtration only) for the 30 cm body phantom with calcium and iodine inserts and 80/140 kV. The error bars on the experimental plots indicate standard deviation of the CT numbers within the inserts. Linear regressions were used to determine the slopes. The ratios of the slopes were calculated and defined as the DE_{ratio} . For example, the DE_{ratio} of iodine for the 0.8 mm of tin filtration is equal to $41.87/12.34 = 3.39$.

TABLE IV. The DE ratio difference between two clinically relevant materials (calcium-iodine and calcium-iron) for the factory-installed filtration and the added high-energy beam filtration with tin.

	Body 80/140 kV					Large body 100/140 kV		
	Factory-installed filtration	Added filtration (tin 0.8 mm)	Increase (%)	Added filtration (tin 0.5 mm)	Increase (%)	Factory-installed filtration	Added filtration (tin 0.5 mm)	Increase (%)
Ca and I	0.39	1.48	279	1.15	195	0.23	0.76	230
Ca and Fe	0.28	1.02	264	0.86	207	0.13	0.51	292

energy spectrum. Currently, the separation between the 100 and 140 kV spectra is too small to produce sufficient DE contrast. Using the filtration optimized for the 100/140 kV imaging, the amount of DE contrast between calcium and iodine can be increased from 0.23 (factory filtration, 100/140 kV) to 0.76 (230%) for a 40 cm wide patient. This will result in 95% (0.76/0.39) higher DE contrast between calcium and iodine than can currently be achieved for a 30 cm wide patient using 80/140 kV and the factory-installed filtration (0.39).

Our study had the following limitations. First, the experimental validation of the simulated DE ratios presented in our study was limited to the factory filtration. Experimental measurements with the proposed added filtration would require scanner modifications in order to place the added filter material securely enough to allow gantry rotation. This is needed to perform a CT acquisition such that image noise can be evaluated; experiments using a stationary tube cannot address CT image noise. However, we believe that validating our simulations for the factory filtration is sufficient to show that the software used produced accurate results. The relatively small but consistent difference of 6% between the experimental and simulated DE_{ratio} values likely resulted from the fact that our reconstruction kernel and beam-hardening corrections were not exactly the same as those implemented on the DSCT system. However, it is not the absolute DE_{ratio} values but rather the difference between two materials that is of primary importance for DE material characterization. Experimentally, the differences between the measured DE_{ratio} values for calcium and iodine were in excellent agreement for two different simulations (both 30 and 40 cm virtual thorax phantom).

Second, we did not explicitly investigate how the filter choice affected radiation dose. However, it follows directly from Eq. (6) that maximization of the difference in the DE ratios simultaneously minimizes noise in the material-specific image. This ensures that the minimum radiation dose can be used to accomplish the specific material differentiation task. With knowledge of the optimum additional filtration for the high-energy beam, modification of a DSCT system is under way in order to evaluate the effect of additional filtration on both dose (in phantoms) and overall image quality (in animals).

V. CONCLUSIONS

We have shown that the use of added filtration for the high-energy x-ray tube of a DSCT system operated in the DE

mode can dramatically increase the DE contrast between clinically relevant materials. Although seven single-element materials were found to perform similarly well at proper thicknesses, tin is an ideal filter material, as it is inexpensive and easy to machine. Appropriate thicknesses were found to be 0.5 or 0.8 mm, respectively, for the large and normal size patient attenuations. This increase in the DE contrast should significantly improve the performance of DE material discrimination algorithms, increasing the clinical value of existing DE applications and opening possibilities for new advanced clinical applications. Furthermore, added filtration can be used with 100/140 kV tube potentials to allow improved DE imaging of large patients, who currently have reduced quality of DE exams due to unacceptable noise levels at 80 kV.

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