Published in final edited form as:

Phys Med Biol. ; 64(11): 115019. doi:10.1088/1361-6560/ab1c34.

Dual-source photon counting detector CT with a tin filter: a phantom study on iodine quantification performance

Ashley Tao^{1,2}, Richard Huang^{1,3}, Shengzhen Tao¹, Gregory J Michalak¹, Cynthia H McCollough¹, and Shuai Leng^{1,4}

¹Department of Radiology, Mayo Clinic, 200 First Street SW, Rochester, MN 55905, United States of America

²Current institution: Gundersen Health System, La Crosse, WI 54601, United States of America

³Current institution: University of Southern California, Los Angeles, CA 90007, United States of America

Abstract

Photon counting detectors (PCD) can provide spectral information to enable iodine quantification through multi-energy imaging but performance is limited by current PCD technology. The purpose of this work is to evaluate iodine quantification in a phantom study using dual-source PCD-CT (DS-PCD-CT), and compare to single-source (SS)-PCD-CT and traditional DS energy integrating detector (EID)-based dual-energy CT.

A multi-energy CT phantom with iodine inserts (0 to 15 mg ml⁻¹ concentration) was imaged on a research SS-PCD-CT scanner (CTDI_{vol} = 18 mGy). A DS-PCD-CT was emulated by acquiring two sequential scans (CTDI_{vol} = 9 mGy each) using tube potentials: 140 kVp/80 kVp, 140 kVp/100 kVp and 140 kVp/120 kVp. For each kVp, 1 or 2 energy bins were reconstructed to achieve either dual-energy or quadruple energy CT. In addition to these energy combinations, a Sn filter was used for the high tube potential (140 kVp) of each kVp pair. For comparison, the same phantom was also scanned on a commercially available DS-EID-CT with matched radiation dose (CTDI_{vol} = 18 mGy). Material decomposition was performed in image space using a standard least-squares based approach to generate iodine and water-specific images. The root-mean-square-error (RMSE) measured over each insert from the iodine image was used to determine iodine accuracy.

The iodine RMSE from SS-PCD (140 kVp with 2 energy bins) was 2.72 mg ml⁻¹. The use of a DS configuration with 1 energy bin per kVp (140 kVp/80 kVp) resulted in a RMSE of 2.29 mg ml⁻¹. Two energy bins per kVp further reduced iodine RMSE to 1.83 mg ml⁻¹. The addition of a Sn filter to the latter quadruple energy mode reduced RMSE to 1.48 mg ml⁻¹. RMSE for DS-PCD-CT (2 energy bins per kVp) decreased by 1.3% (Sn140 kVp/80 kVp) and 15% (Sn140 kVp/100 kVp) as compared to DS-EID-CT.

DS-PCD-CT with a Sn filter improved iodine quantification as compared to both SS-PCD-CT and DS-EID-CT.

⁴ Author to whom any correspondence should be addressed. leng.shuai@mayo.edu.

Keywords

multi-energy CT; photon counting detector CT; iodine quantification; dual-source CT; spectral separation; tin filter

Introduction

There has been recent interest in photon counting detector CT (PCD-CT) due to advancements in solid state detector technology and faster application specific integrated circuits (ASICs) (Schlomka *et al* 2008, Shikhaliev 2008, Iwanczyk *et al* 2009, Anderson *et al* 2010, Kappler *et al* 2010, 2012, Xu *et al* 2012, Taguchi and Iwanczyk 2013, Muenzel *et al* 2017, Dangelmaier *et al* 2018). Since the introduction of a research whole-body PCD-CT scanner, a number of studies have shown that PCD-CT improves iodine contrast to noise ratio (Gutjahr *et al* 2016, Yu *et al* 2016a), decreases the effect of electronic noise, reduces metal artifacts (Zhou *et al* 2018) and improves dose efficiency for high resolution imaging (Leng *et al* 2016, 2018) as compared to traditional energy integrating detector (EID) CT scanners. The primary difference between these detectors is the ability of PCDs to discriminate photons based on their energies. Photons can be divided into any number of bins where photons above a defined threshold are binned together. The number of energy bins is only limited by the number of thresholds allowed for a specific system. Depending on the detector configuration, either two or four different energy thresholds can be selected for the PCD-CT scanner investigated in this study (Yu *et al* 2016a). Subtraction of adjacent energy thresholds can generate images with various energy bins.

One application that can benefit from PCD-CT's energy discrimination capabilities is tissue characterization using multi-energy CT data (McCollough *et al* 2015), in which the composition of tissues are derived from the x-ray attenuation coefficients measured at a minimum of two different energies. Early work focused on developing proof of concept for multi-energy imaging using a small field of view on small animal PCD based CT scanners (Schlomka *et al* 2008, Shikhaliev 2008, Anderson *et al* 2010, Wang *et al* 2011, Muenzel *et al* 2017, Dangelmaier *et al* 2018). The introduction of a whole-body PCD-CT has since facilitated the ability to perform quantitative imaging in adult-sized phantoms and patients, such as iodine quantification, urinary and kidney stone characterization (Leng *et al* 2017, Ferrero *et al* 2018, Marcus *et al* 2018).

EID-CT, on the other hand, is not capable of energy discrimination. However, there are several approaches that are clinically used to provide image data at two different energies, e.g. fast kVp switching (Xu *et al* 2009), sandwich detector (Carmi *et al* 2005) and a dual source (DS) configuration (Flohr *et al* 2006). One of the advantages of using a DS-CT is the ability to add a Sn filter. The use of a Sn filter hardens the higher kVp beam (typically 140 kVp or 150 kVp) by selectively absorbing low energy photons, thereby increasing the energy separation between low and high kVp images (Primak *et al* 2010).

Although multi-energy imaging can be performed on a single source (SS)-PCD, a simulation study by Faby *et al* showed that SS-PCD might not perform as well as DS-EID-CT due to limited performance of PCDs caused in part by non-ideal detector response such as charge

sharing and K-escape x-rays (Faby *et al* 2015). For an ideal PCD, a single keV threshold would be used to separate photons from an x-ray spectrum perfectly into either the higher energy bin or lower energy bin, as shown in figure 1(a). However, energy detection is limited by the finite energy resolution of the PCD detector, and the detector inherently suffers from incomplete charge collection from charge sharing, cross talk and trapping, resulting in a low energy tail further degrading energy resolution. Based on a Monte Carlo simulation for a research whole-body PCD-CT system, the expected energy spectrum for a realistic PCD with two energy bins would appear as figure 1(b). Although this is still sufficient to provide dual-energy images, its performance is substantially degraded due to the significant spectral overlap between the two energy bins. The simulation study by Faby *et al* showed that an EID-based DS dual-energy configuration with Sn filter provided better performance than SS-PCD-CT without a Sn filter. That study also simulated a hypothetical DS-PCD scanner with two tubes operated at different tube potentials (kVps), and its results suggested that the DS-PCD-CT might perform better than SS-PCD-CT and DS-EID-CT systems (Faby *et al* 2015).

Therefore, the purpose of this study is to experimentally assess the performance of DS-PCD-CT in iodine quantification based on a whole-body SS-PCD-CT. Since a DS-PCD-CT system does not currently exist, a SSPCD-CT was used to emulate a dual-source configuration with sequentially acquired phantom scans at two different tube potentials. Our main comparison in multi-energy imaging is between SS-PCD-CT and DS-PCD-CT. In addition, DS-PCD-CT was compared to a commercially available DS-EID-CT. For all comparisons, the total radiation dose was matched among DS-EID-CT, SS-PCD-CT and DS-PCD-CT.

Methods

Research photon counting detector CT scanner

The research PCD-CT scanner used in this study was based on the same platform as a second generation, DS-CT scanner (SOMATOM Definition Flash, Siemens Healthineers, Forchheim, Germany) with one tube/detector pair corresponding to an EID with a 50 cm field of view (FOV) and the second tube/detector pair corresponding to a CdTe-based PCD with a 27.5 cm FOV (Yu *et al* 2015). Due to the limited FOV of the PCD-CT, a data completion scan (DCS) is required when imaging subjects larger than 27.5 cm to avoid truncation artifacts (Yu *et al* 2016b). The DCS is performed by scanning the subject on the EID subsystem either before or after the PCD-CT scan and the data are used to estimate the missing projection data beyond the scan FOV of the PCD system. The purpose is solely to use data from the larger FOV EID to prevent truncation artifacts and does not provide additional PCD information beyond the maximum 27.5 cm FOV, nor does it impact image quality, e.g. noise inside the FOV (Yu *et al* 2016b).

The research PCD-CT scanner has four different acquisition modes: macro, chess, sharp, and ultra-high resolution. For the purposes of this study, which is to investigate multi-energy performance of PCD using a dual-source approach, the scanner was operated in macro mode. In this mode, native detector pixels (0.225 mm) are grouped into 4×4 arrays to generate 0.9 mm effective detector pixels, resulting in 0.5 mm resolution at iso-center given

a 1.8 magnification. Additional details of the PCD-CT system can be found in other publications (Yu *et al* 2015, Leng *et al* 2018).

Two energy thresholds, low (T_L) and high (T_H), can be selected, resulting in spectral information with 4 image types, 3 of which are unique. The first two images correspond to the threshold images, where photons greater than the low and high thresholds are included in the dataset up to the prescribed tube potential generating threshold low (T_L) and threshold high (T_H) images respectively. The second two images correspond to the subtraction of adjacent energy thresholds. The T_H image is subtracted from the T_L image to generate an image between T_L and T_H (i.e. Bin 1). The second energy bin image (i.e. Bin 2) corresponds to energy level between T_H and the maximum energy and is equivalent to the T_H image.

Phantom

A multi-energy CT phantom (40 cm × 30 cm × 16.5 cm) (Sun Nuclear, Middleton, WI) was used to simulate the attenuation of a medium to large sized patient. The phantom consisted of solid water and five inserts (28.5 mm diameter) with iodine concentrations of 0, 2, 5, 10, and 15 mg ml⁻¹ arranged as shown in figure 2. All 5 inserts were within the 27.5 cm scan FOV of the PCD-CT. Several other inserts were used solely as placeholders and were not included in the data analysis.

PCD-CT image acquisition

Two main system configurations were investigated with the PCD: SS-PCD-CT and DS-PCD-CT. The SS-PCDCT served as a baseline reference, which was performed at 140 kVp with 2 energy thresholds ($T_L = 25$ keV and $T_H = 69$ keV) and a CTDI_{vol} of 18 mGy. The system's maximum tube potential of 140 kVp was selected to provide the best spectral separation (a higher tube potential allows for greater difference between energy bins due to the non-ideal low energy tail). The thresholds were determined through Monte-Carlo simulation such that bins 1 and 2 resulted in approximately equal number of photons (i.e. 50% of the total number of photons are allocated to each bin).

Since the research scanner is only equipped with a SS-PCD, a DS-PCD-CT system was emulated with two separate acquisitions using the SS-PCD-CT with different tube potentials (kVp pairs). For DS-PCD-CT, tube potential for one of the scans was fixed at 140 kVp, while the other scan was performed at tube potentials of 80, 100, or 120 kVp. Additional scans were performed with a 0.4 mm Sn filter added to the 140 kVp. A Sn filter was used only for the DS-PCD-CT configuration and not SS-PCD-CT because the Sn filter shifts the mean energy of the spectrum towards a higher energy thereby decreasing the separation between the mean energies of the two acquired energy bins. Each scan was performed at half the CTDI_{vol} of the SS-PCD-CT (18 mGy), i.e. 9 mGy, by adjusting tube current accordingly. This was to match total radiation dose between SS-PCD-CT and DS-PCDCT. For DS-PCD-CT, two different multi-energy configurations were investigated for each tube potential pair. In the first configuration, PCD data acquired at each kVp was treated as a whole, e.g. using T_L data only, which resulted in a dual energy CT configuration very similar as conventional DS-DECT, except using PCD instead of EID. In the second configuration, two energy bin data sets (e.g. bin 1 and bin 2) from each kVp scan were used, which resulted in a total of

four data sets, i.e. quadruple energy CT. Table 1 summarizes all the different kVp and energy bin combinations considered in this study.

The energy thresholds were determined similarly as described for the SS-PCD-CT such that approximately 50% of the photons were distributed in bin 1 and 50% of the photons were distributed in bin 2 for each respective kVp. The thresholds used for SS-PCD-CT and DS-PCD-CT are summarized with the corresponding mean energy in table 2. The mean energies were calculated based on equation (1), where E_0 denotes the mean energy of the energy bin of interest; $s(E)$ is the portion of the transmitted spectrum for the corresponding energy bin, with consideration of detector response function that was calculated from MC simulations.

$$E_0 = \frac{\int E \cdot s(E) dE}{\int s(E) dE}. \quad (1)$$

For all PCD-CT scans, images were reconstructed using standard filtered back projection with a quantitative medium smooth kernel (D30), 5 mm slice thickness, 27.5 cm FOV and a matrix size 512×512 .

EID-CT image acquisition

For comparison with commercial EID dual energy systems, a dual-source, dual-energy CT scanner (SOMATOM Definition Flash, Siemens Healthineers, Forchheim, Germany) was used, the same platform upon which the PCD-CT system was built, ensuring that the system geometry and x-ray source were equivalent. DS-EID-CT corresponds to the traditional DE technique with clinically used tube potential combinations (Sn 140 kVp/80 kVp and Sn 140 kVp/100 kVp). These techniques used a default dose partition of 0.7:1 for Sn 140 kVp/80 kVp and 0.67:1 for Sn 140 kVp/100 kVp that cannot be changed by users (measured from the scanner). A total CTDI_{vol} of 18 mGy was used for each kVp pair, which was the same as that of the PCD acquisitions. Images were also reconstructed using standard filtered back projection with a D30 kernel, 5 mm slice thickness, 27.5 cm FOV, and a matrix size of 512×512 .

Material decomposition and data analysis

Material decomposition was performed in image space using the least squares method with images acquired using SS-PCD, DS-EID, and DS-PCD-CT configurations to generate iodine and water-specific images. Denoting ω_i as the mass density of the i th basis material for each pixel, the effective linear attenuation measurements from a multi-energy CT acquisition can be expressed as follows:

$$\mu_e = \sum_{i=1}^M \left(\frac{\mu}{\rho} \right)_{e,i} \omega_i \quad (2)$$

where μ_e denotes the effective linear attenuation coefficients of the e th energy channel ($1 \leq e \leq E$) for the current pixel of interest; $\left(\frac{\mu}{\rho}\right)_{e,i}$ is the mass attenuation coefficient of the i th material ($1 \leq i \leq M$) and the e th, energy channel, which can be obtained from a calibration scan with known material composition, and M, E represent the number of basis materials and number of energy spectra settings for a multi-energy CT acquisition. Here, we focus on iodine and water-based decomposition, hence $M=2$. The effective linear attenuation coefficients were determined with the mean CT number of the e th energy channel $\mu = \frac{CT+1000}{1000} * \mu_{water}$, where μ_{water} is the attenuation coefficient of water, which is calibrated to be ($\mu_{water} = 0.1907 \text{ cm}^{-1}$) for this scanner (Chen *et al* 2015). Equation (2) can be expressed into a linear algebraic form:

$$\boldsymbol{\mu} = \mathbf{M}\boldsymbol{\omega} \quad (3)$$

where $\boldsymbol{\omega} = [\omega_1, \omega_2]^T$, $M(e, i) = \left(\frac{\mu}{\rho}\right)_{e,i}$, $\boldsymbol{\mu} = [\mu_1, \mu_2, \dots, \mu_E]^T$.

The standard image-domain basis material decomposition based on least-square fitting calculates the material specific images, which can be written as a matrix inversion process:

$$\boldsymbol{\omega} = (\mathbf{M}^T \mathbf{M})^{-1} \mathbf{M}^T \boldsymbol{\mu}. \quad (4)$$

The number of energy spectra settings, E , is determined based on the data acquisition setup. For a SS-PCD with two energy thresholds, as well as the conventional DS-EID-CT, $E=2$. For the DS-PCD-CT, $E=2$ when the T_L dataset of each source is used, while $E=4$ when the two bin data sets of each source are used.

The iodine concentration was then measured from the output iodine-specific images. The mean iodine concentration ($I_{measured}$) was measured with a 2.5 cm circular ROI for each of the iodine inserts. The root mean square error (RMSE) was calculated based on the known iodine concentration of each insert (I_{true}) and used as the figure of merit to quantify the accuracy and precision of each of the kVp and energy bin combinations (equation (5)). The RMSE considers all pixels, N_{pixels} for a given iodine insert, all five inserts $N_{inserts}$ and averaged over the central 10 slices of the phantom ($N_{slices}=10$). Note that the RMSE considers both bias and noise in the estimation.

$$RMSE = \sqrt{\frac{1}{N_{slices} \times N_{inserts} \times N_{pixels}} \sum_{slice=1}^{N_{slices}} \sum_{n=1}^{N_{inserts}} \sum_{p=1}^{N_{pixels}} (I_{true} - I_{measured})^2}. \quad (5)$$

Since the RMSE is a single figure of merit which considers the error from all iodine inserts, a summary of the bias from each iodine insert was also included where the bias is the difference between the measured mean iodine concentration and true iodine concentration. A two-way ANOVA test was performed on the bias data to determine if there was a significant

difference in iodine quantification bias between different insert concentrations. In addition, a paired sample *t*-test was performed between the RMSE of DS-PCD-CT and SSPCD-CT, and between that of DS-PCD-CT and DS-EID-CT to determine if there was a significant difference between these system configurations.

Results

The SS-PCD-CT result was used as the reference for comparison with DS-PCD-CT. The iodine quantification RMSE for SS-PCD-CT was 2.72 mg ml^{-1} using two energy bins and 140 kVp. An example slice through the center of the phantom is shown in figure 3(a). The RMSE for the dual-source approach with 1 energy bin per kVp (dual-energy PCD-CT, 1 energy bin per source) using the 140 kVp/80 kVp pair was 2.29 mg ml^{-1} (figure 3(b)), a 16% reduction compared to the reference SS-PCD-CT. Using the same kVp pair but with 2 energy bins kVp⁻¹ (quadruple-energy PCD-CT, 2 energy bins per source) further decreased RMSE to 1.83 mg ml^{-1} (figure 3(c)), a 33% reduction in RMSE from the reference SS-PCD-CT configuration. Using the same kVp and energy bin combination, with the addition of a Sn filter on the high kVp data, further decreased the RMSE to 1.49 mg ml^{-1} (figure 3(d)), a 46% reduction in RMSE from the reference SS-PCD-CT configuration.

Figure 4 summarizes the RMSE for SS-PCD-CT and all the different kVp and energy combinations for DSPCD-CT listed in table 1. With reference to the SS-PCD-CT, RMSE for the DS-PCD-CT configuration with 2 energy bins (1 bin for each kVp) decreased only for the 140 kVp/80 kVp pair but increased for the 140 kVp/100 kVp and 140 kVp/120 kVp pair. The use of quadruple energy bins (2 energy bins for each kVp) reduced RMSE for all kVp pairs relative to the reference SS-PCD-CT. The addition of a Sn filter further reduced RMSE for all kVp pairs. In general, RMSE decreased as energy separation increased between kVp pairs. All DS-PCD-CT kVp pairs with quadruple energy bins outperformed SS-PCD-CT, with or without a Sn filter (p -value = 1×10^{-14} (with Sn filter), p -value = 6×10^{-14} (without Sn filter)). The best performance came from 140 kVp/80 kVp with a tin filter and 2 energy bins for each kVp (quadruple energy).

The individual bias for each insert and kVp combination is summarized in figure 5. The bias for PCD-CT acquisitions for all kVp combinations (both SS and DS) was less than 0.5 mg I ml^{-1} except for the 140 kVp/120 kVp pair with 1 energy bin for each kVp. A two-way ANOVA test showed there was not a significant difference in the bias magnitude between each iodine insert concentration (p -value = 0.26).

For comparison of DS-PCD-CT with DS-EID-CT, an example slice from the center of the phantom along with the iodine quantification RMSE are summarized in figure 6 for Sn 140 kVp/80 kVp and Sn 140 kVp/100 kVp. For the Sn 140 kVp/80 kVp pair, RMSE decreased by 1.3% (from 1.5 mg I ml^{-1} to $1.48 \text{ mg I ml}^{-1}$) using DS-PCD-CT compared to DS-EID-CT. RMSE decreased by 15% (from $2.31 \text{ mg I ml}^{-1}$ to $1.96 \text{ mg I ml}^{-1}$) for the Sn 140 kVp/100 kVp pair using DS-PCD-CT compared to DS-EID-CT. A paired sample *t*-test between the RMSE from DS-PCD-CT and DS-EID-CT with the Sn 140 kVp/80 kVp pair resulted in p -value = 0.73 suggesting there was not a significant difference between the two

system configurations. However, for the Sn 140 kVp/100 kVp pair, the p -value = 4×10^{-13} indicating there was a significant difference in RMSE between PCD and EID.

Discussion

While a SS-PCD-CT can provide spectral information to perform DECT, this phantom study showed that a dual source configuration with two energy bins per tube potential and a Sn filter provided better iodine quantification for all kVp pairs investigated. The RMSE decreased when kVp pairs had greater spectral separation (i.e. 140 kVp/80 kVp as compared to 140 kVp/120 kVp). The addition of 2 energy bins for each kVp provided additional spectral information resulting in a lower RMSE. Finally, the addition of a Sn filter on the high kVp spectrum selectively attenuated low energy photons, thereby increasing the spectral separation, resulting in further decreased RMSE.

In comparison to DS-EID-CT, there was only a small improvement in RMSE using DS-PCD-CT for the Sn 140 kVp/80 kVp pair (RMSE decreased by 1.3%). However, there was a significant improvement for the Sn 140 kVp/100 kVp pair (RMSE decreased by 15%). For many DECT systems (e.g. GE DECT based on fast kVp switching), scans are typically performed with 140 kVp/80 kVp pairs. On DSDE CT, Sn 140 kVp/80 kVp is also available which can improve spectral separation. However, both 140 kVp/80 kVp and Sn 140 kVp/80 kVp pairs are limited for very large patients due to photon starvation. For DSDE CT, other kVp pairs are available and studies have shown that Sn 140 kVp/100 kVp can be used to image larger patients (Primak *et al* 2010, Siegel *et al* 2013). As can be seen from the RMSE summary in figure 4, the relative improvement in RMSE for DS-PCD-CT with Sn 140 kVp/100 kVp was greater than Sn 140 kVp/80 kVp. This suggests that DS-PCD may be more beneficial for larger patients. In addition, PCD-CT has been shown to be less susceptible to beam hardening artifacts and has reduced electronic noise (Yu *et al* 2015, 2016a, Pourmorteza *et al* 2017), which may also benefit large patient imaging.

The RMSE from DS-PCD-CT with Sn 140 kVp/120 kVp and quadruple energy scan (2 energy bins kVp^{-1}) was comparable to the reference SS-PCD-CT. The Sn 140 kVp/120 kVp pair is not encountered in commercial DS-EID-CT due to the significant spectral overlap. However, given the greater improvement in quantification using tube potential pairs with less spectral separation (DS-PCD-CT), it may be feasible to image very large patients using Sn 140 kVp/120 kVp. This would be beneficial for very large patients because multi-energy imaging requires spectral separation but images from the lower tube potential are overwhelmed by noise from photon starvation.

In comparison to the simulation study performed by Faby *et al*, there were a number of similarities in our results, despite the many differences in experimental technique (i.e. kVp combinations, energy thresholds, energy bins, material decomposition algorithm, phantom versus simulation data, image reconstruction and figure of merit). In this work, we focused on a phantom study using a scanner which represents the performance of the PCD technology that is currently available. In general, our results were in agreement with the simulation study by Faby *et al* comparing similar configurations/techniques (e.g. SS-PCD-CT versus DS-PCD-CT, number of energy bins) (Faby *et al* 2015). Faby *et al*'s figure of

merit for their comparisons was the % noise in the image relative to a DS-EID-CT with tube potentials of Sn 140 kVp/100 kVp. Both our phantom study (15% increase in RMSE) and Faby *et al*'s simulation study (1% increase in noise) showed SS-PCD-CT did not outperform DS-EID-CT due to the non-ideal properties of PCDs (e.g. charge sharing, K-escape x-rays). In addition, DS-PCD-CT with a Sn filter showed improvement in performance as compared to DS-EID-CT for both studies (21% noise reduction in Faby's simulation study, or 15% reduction in RMSE in our phantom study). It should be noted that the absolute measurements cannot be directly compared due to differences in experimental techniques. However, the general relationship observed between the phantom and simulation studies were in agreement comparing DS-EID-CT, SS-PCD-CT and DS-PCD-CT.

In this study, standard filtered back projection (FBP) was used as the reconstruction algorithm. This is because the focus of this work was to compare the system performance between SS-PCD and DS-PCD; other confounding factors were purposely excluded. Therefore, linear FBP was used, as opposed to iterative reconstruction to avoid the introduction of any non-linear effects. For the same reason, the simplest image-based, least square fitting material decomposition method was selected. Results shown in this study therefore demonstrate the fundamental performance of SS-PCD and DS-PCD systems.

The proposed technique can be implemented using a dual-source approach, or a dual-scan approach as was performed in this study. The advantage of dual-scan approach is that it does not require a 2nd photon counting detector and potentially could be achieved with currently available hardware. However, patient motion between the two scans could be a concern for the latter approach and sophisticated registration software is required. A dual-source configuration is preferred as it is much less affected by patient motion, with the two data sets acquired simultaneously.

One limitation of our study was the selection of the filter parameters to improve spectral separation. The PCDCT scanner was equipped only with a 0.4 mm Sn filter. Optimal filter material/thickness for PCD-CT was not investigated in this study. Another limitation of our study was that some of the input parameters to our DS-PCD configuration were not optimized, such as dose partitioning and selection of energy threshold values (partitioning within the spectra for each tube potential). As a first step for proof of concept, a 50%/50% dose partition was selected for all DS-PCD configurations, and the energy thresholds were selected to provide equal number of photons for each energy bin. Both were not necessarily the optimal settings for the iodine quantification task but were selected mainly for simplicity. It is anticipated that iodine quantification of DS-PCD can be further improved with optimization of these parameters, which our team is actively pursuing. In contrast, the DS-EID-CT used the vendor optimized dose partition between the two tube potentials, which gave an advantage to the DS-EID-CT results used to compare to PCD results. Therefore, the improvement of DS-PCD-CT compared to DS-EID-CT is expected to be higher than that observed in this study, once the DS-PCD-CT settings are optimized.

Conclusion

A dual-source PCD-CT with 2 energy bins per tube potential and a Sn filter improved iodine quantification performance by as much as 46% compared to SS-PCD-CT in a phantom study. This was due to increased spectral separation contributing from the energy discrimination of PCD, using a dual-source acquisition approach, and adding a Sn filter for the high tube potential spectrum. The Sn 140 kVp/80 kVp pair using the DS-PCD-CT configuration resulted in the lowest iodine quantification RMSE.

Acknowledgments

The project described was supported by Grant numbers EB16966 and C06 RR018898 from the National Institute of Health. The content is solely the responsibility of the authors and does not necessarily represent the official views of the National Institute of Health.

Part of this work was presented at RSNA 2018 annual meeting.

Conflicts of interest

Dr Cynthia McCollough received industry grant funding from Siemens Healthineers. No other authors have any conflicts to disclose.

References

- Anderson NG et al. 2010 Spectroscopic (multi-energy) CT distinguishes iodine and barium contrast material in MICE Eur. Radiol 20 2126–34 [PubMed: 20309554]
- Carmi R, Naveh G and Altman A 2005 Material separation with dual-layer CT IEEE Nuclear Science Symp. Conf. Record vol 4, pp 1876–8
- Chen B, Leng S, Yu L, Yu Z, Ma C and McCollough C 2015 Lesion insertion in the projection domain: methods and initial results Med. Phys 42 7034–42 [PubMed: 26632058]
- Dangelmaier J et al. 2018 Experimental feasibility of spectral photon-counting computed tomography with two contrast agents for the detection of endoleaks following endovascular aortic repair Eur. Radiol 28 3318–25 [PubMed: 29460069]
- Faby S et al. 2015 Performance of today's dual energy CT and future multi energy CT in virtual non-contrast imaging and in iodine quantification: a simulation study Med. Phys 42 4349–66 [PubMed: 26133632]
- Ferrero A, Gutjahr R, Halaweish AF, Leng S and McCollough CH 2018 Characterization of urinary stone composition by use of whole-body, photon-counting detector CT Acad. Radiol 25 1270–6 [PubMed: 29454545]
- Flohr TG et al. 2006 First performance evaluation of a dual-source CT (DSCT) system Eur. Radiol 16 256–68 [PubMed: 16341833]
- Gutjahr R et al. 2016 Human imaging with photon counting-based computed tomography at clinical dose levels: contrast-to-noise ratio and Cadaver studies Investigative Radiol. 51 421–9
- Iwanczyk JS et al. 2009 Photon counting energy dispersive detector arrays for x-ray imaging IEEE Trans. Nucl. Sci 56 535–42 [PubMed: 19920884]
- Kappler S. et al. 2012; First results from a hybrid prototype CT scanner for exploring benefits of quantum-counting in clinical CT. SPIE Med. Imaging. 8313:11.
- Kappler S, Glasser F, Janssen S, Kraft E and Reinwand M 2010 A research prototype system for quantum-counting clinical CT Proc. SPIE 7622 76221Z
- Leng S. et al. 2016; Dose-efficient ultrahigh-resolution scan mode using a photon counting detector computed tomography system. J. Med. Imaging. 3:043504.
- Leng S et al. 2017 Spectral performance of a whole-body research photon counting detector CT: quantitative accuracy in derived image sets Phys. Med. Biol 62 7216–32 [PubMed: 28726669]

- Leng S et al. 2018 150-um spatial resolution using photon-counting detector computed tomography technology: technical performance and first patient images *Investigative Radiol* 53 655–62
- Marcus RP et al. 2018 Detection and characterization of renal stones by using photon-counting-based CT *Radiology* 289 436–42 [PubMed: 30084728]
- McCollough CH, Leng S, Yu L and Fletcher JG 2015 Dual- and multi-energy CT: principles, technical approaches, and clinical applications *Radiology* 276 637–53 [PubMed: 26302388]
- Muenzel D et al. 2017 Spectral photon-counting CT: initial experience with dual-contrast agent K-edge colonography *Radiology* 283 723–8 [PubMed: 27918709]
- Pourmorteza A et al. 2017 Photon-counting CT of the brain: in vivo human results and image-quality assessment *AJNR Am. J. Neuroradiol* 38 2257–63 [PubMed: 28982793]
- Primak AN et al. 2010 Dual-source dual-energy CT with additional tin filtration: dose and image quality evaluation in phantoms and in vivo *AJR Am. J. Roentgenol* 195 1164–74 [PubMed: 20966323]
- Schlomka JP et al. 2008 Experimental feasibility of multi-energy photon-counting K-edge imaging in pre-clinical computed tomography *Phys. Med. Biol* 53 4031–47 [PubMed: 18612175]
- Shikhaliev PM 2008 Energy-resolved computed tomography: first experimental results *Phys. Med. Biol* 53 5595–613 [PubMed: 18799830]
- Siegel M, Ramirez-Giraldo J and Graser A 2013 Abdominal dual-source dual-energy CT: uses in clinical practice *Appl. Radiol* 42 10–6
- Taguchi K and Iwanczyk JS 2013 Vision 20/20: single photon counting x-ray detectors in medical imaging *Med. Phys* 40 100901 [PubMed: 24089889]
- Wang X, Meier D, Taguchi K, Wagenaar DJ, Patt BE and Frey EC 2011 Material separation in x-ray CT with energy resolved photon-counting detectors *Med. Phys* 38 1534–46 [PubMed: 21520865]
- Xu D. et al. 2009; Dual energy CT via fast kVp switching spectrum estimation. *Proc. SPIE*. 7258:72583T.
- Xu Q et al. 2012 Image reconstruction for hybrid true-color micro-CT *IEEE Trans. Biomed. Eng* 59 1711–9 [PubMed: 22481806]
- Yu Z. et al. 2015; Initial results from a prototype whole-body photon-counting computed tomography system. *Proc. SPIE*. 9412:94120W.
- Yu Z et al. 2016a Evaluation of conventional imaging performance in a research whole-body CT system with a photon-counting detector array *Phys. Med. Biol* 61 1572–95 [PubMed: 26835839]
- Yu Z et al. 2016b How low can we go in radiation dose for the data-completion scan on a research whole-body photon-counting computed tomography system *J. Comput. Assist. Tomogr* 40 663–70 [PubMed: 27096399]
- Zhou W et al. 2018 Reduction of metal artifacts and improvement in dose efficiency using photon-counting detector computed tomography and tin filtration *Investigative Radiol* 54 204–11

Figure 1. Simulation of 140 kVp spectra with two energy bins for: (a) an ideal PCD and (b) a realistic PCD on the research PCD-CT system.

Figure 2. Solid water based multi-energy CT phantom with iodine inserts identified by red circles (center: 0 mg I ml⁻¹, top: 2 mg I ml⁻¹, left: 5 mg I ml⁻¹, bottom: 10 mg I ml⁻¹, right: 15 mg I ml⁻¹).

Figure 3.

Iodine-specific images using (a) SS-PCD-CT –140 kVp (2) where the number in parentheses represents the number of energy bins per kVp (b) DS-PCD-CT 140 kVp/80 kVp (1 energy bin for each kVp, i.e. using the energy threshold low (T_L) data of each kVp) (c) DS-PCD-CT 140 kVp/80 kVp (2 energy bins for each kVp) (d) DS-PCD-CT 140 kVp/80 kVp (2 energy bins for each kVp) with a tin filter applied to the 140 kVp acquisition.

Figure 4. Summary of iodine quantification RMSE comparing SS-PCD and DS-PCD with different kVp pairs.

Figure 5.
Iodine quantification bias for each insert and kVp combination.

Figure 6. Iodine specific images generated using images from (a) EID Sn 140 kVp/80 kVp (b) EID Sn 140 kVp/100 kVp (c) PCD Sn 140 kVp/80 kVp (d) PCD Sn 140 kVp/100 kVp.

Table 1.

Summary of tube potentials, $CTDI_{vol}$ (per scan and total) and energy bin combinations for SS-PCD-CT and DS-PCD-CT. Sn 140 indicates a tin filter was added to the 140 kVp beam.

	Tube potential (kVp) combination	$CTDI_{vol}$ (mGy)	# energy bins/tube potential	
SS-PCD-CT	140	18	2	
DS-PCD-CT	140/80	9/9 (18 total)	1/1	2/2
	Sn 140/80	9/9 (18 total)	1/1	2/2
	140/100	9/9 (18 total)	1/1	2/2
	Sn 140/ 100	9/9 (18 total)	1/1	2/2
	140/120	9/9 (18 total)	1/1	2/2
	Sn 140/120	9/9 (18 total)	1/1	2/2

Table 2.

Energy thresholds used for each tube potential (kVp) and corresponding mean energy (E_0) for each energy bin.

Tube potential (kVp)	T_L (keV)	T_H (keV)	Mean energy (keV)		
			T_L	Bin 1	Bin 2
80	25	54	62	58	67
100	25	61	71	64	79
120	25	65	79	68	90
140	25	69	86	72	100
Sn 140	25	83	97	85	109