Published in final edited form as:

Invest Radiol. 2020 April ; 55(4): 226–232. doi:10.1097/RLI.0000000000000634.

A Universal Protocol for Abdominal CT Examinations Performed on a Photon-Counting Detector CT System:

A Feasibility Study

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Abstract

Objective: The aims of this study were to investigate the feasibility of using a universal abdominal acquisition protocol on a photon-counting detector computed tomography (PCD-CT) system and to compare its performance to that of single-energy (SE) and dual-energy (DE) CT using energy-integrating detectors (EIDs).

Methods: Iodine inserts of various concentrations and sizes were embedded into different sizes of adult abdominal phantoms. Phantoms were scanned on a research PCD-CT and a clinical EID-CT with SE and DE modes. Virtual monoenergetic images (VMIs) were generated from PCD-CT and DE mode of EID-CT. For each image type and phantom size, contrast-to-noise ratio (CNR) was measured for each iodine insert and the area under the receiver operating characteristic curve (AUC) for iodine detectability was calculated using a channelized Hotelling observer. The optimal energy (in kiloelectronvolt) of VMIs was determined separately as the one with highest CNR and the one with the highest AUC. The PCD-CT VMIs at the optimal energy were then compared with DE VMIs and SE images in terms of CNR and AUC.

Results: Virtual monoenergetic image at 50 keV had both the highest CNR and highest AUC for PCD-CT and DECT. For 1.0 mg I/mL iodine and 35 cm phantom, the CNRs of 50 keV VMIs from PCD-CT (2.01 ± 0.67) and DE (1.96 ± 0.52) were significantly higher ($P < 0.001$, Wilcoxon signed-rank test) than SE images (1.11 ± 0.35). The AUC of PCD-CT (0.98 ± 0.01) was comparable to SE (0.98 ± 0.01), and both were slightly lower than DE (0.99 ± 0.01 , $P < 0.01$, Wilcoxon signed-rank test). A similar trend was observed for other phantom sizes and iodine concentrations.

Conclusions: Virtual monoenergetic images at a fixed energy from a universal acquisition protocol on PCD-CT demonstrated higher iodine CNR and comparable iodine detectability than SECT images, and similar performance compared with DE VMIs.

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Supplemental digital contents are available for this article. Direct URL citations appear in the printed text and are provided in the HTML and PDF versions of this article on the journal's Web site (www.investigativeradiology.com).

Keywords

CT; dual-energy CT; photon-counting detector; lesion detectability; virtual monoenergetic images; model observer

In conventional single-energy (SE) CT with energy-integrating detectors (EIDs), tube potential has a significant impact on image quality and radiation dose.^{1–4} For example, a low tube potential (eg, 80 kV) can be used in contrast-enhanced abdominal CT scans to increase iodine signal, which works well for children and small adults. Meanwhile, recent development in CT x-ray tubes with higher power enabled the application of low tube potential on more adult patients.^{5,6} However, for large patients, a low tube potential protocol is not dose efficient due to the lower penetration capability and may not be able to achieve a sufficient number of photons at the detector, resulting in photon starvation artifacts. In this scenario, high tube potentials (eg, 120 or 140 kV) may be more appropriate.⁷ Currently, the appropriate tube potential must be selected based on the patient size and imaging task. This could be achieved using a size-specific technique chart or automatic tube potential selection tools, if available.^{7,8}

Dual-energy (DE) CT is now commercially available and has been shown to provide additional diagnostic information in various clinical scenarios.^{9–13} Similar to SECT, the selection of optimal tube potentials for DE acquisition modes (if available, ie, tube potential pairs on dual-source systems) is dependent on patient size and imaging task.^{14,15} In DECT, virtual monoenergetic images (VMIs) can be generated to simulate CT images acquired with monoenergetic x-ray beams. Virtual monoenergetic images with advanced processing techniques¹⁶ at low energies have been shown to improve iodine contrast and contrast-to-noise ratio (CNR).^{17,18} Alternatively, if the material-specific information that is available from a DECT is not needed, the patient could be scanned with an SE protocol using the optimal tube potential or a DE protocol using the optimal energy VMI. The complexity of these decisions can decrease practice efficiency. More importantly, inappropriate selection of tube potential in SECT, or tube potential pairs in dual-source DECT, can result in suboptimal examinations. A simple and efficient workflow that provides optimal image quality, in terms of iodine CNR and lesion detectability, across different patient sizes is therefore highly desirable.

Recently, research photon-counting detector CT (PCD-CT) systems have become available, and studies have been reported with phantoms,^{19,20} animal models,²¹ cadaveric specimens,²² and patients using clinical dose levels and dose rates.^{23–26} Similarly to DECT, VMIs can be generated from PCD-CT.²⁷ In addition to tube potential and tube current, PCD-CT systems require the selection of additional acquisition parameters, such as energy thresholds, which can further complicate clinical workflow. To simplify clinical decisions, workflow, and protocol management, preliminary data acquired with a research, whole-body PCD-CT system has suggested that the use of VMIs created at a predefined fixed energy from a data set acquired with a predefined fixed protocol (tube potential and energy thresholds) could provide optimal iodine CNR across the range of adult patient sizes.²⁸ However, CNR is

limited in its ability to evaluate reader performance for lesion detection and characterization in the era of nonlinear image reconstruction and processing algorithms.^{29–32}

Therefore, the aims of this study were to determine both lesion CNR and detectability for different adult sizes using fixed energy VMIs acquired with a universal abdominal acquisition protocol and produced with advanced processing techniques on a research whole-body PCD-CT system that is capable of operating at clinically relevant doses, and to compare the results to those of SECT and DECT.

METHODS

We conducted a series of phantom studies to compare the performance of PCD-CT using a universal acquisition protocol with SECT using the optimal tube potential for a given phantom size and dual-source DECT using optimal tube potential pairs for a given phantom size. Two image quality metrics, iodine CNR and low-contrast lesion detectability, were used for these comparisons. For lesion detectability, a channelized Hotelling observer (CHO) model was used, which was previously demonstrated to be correlated to human observer performance in low-contrast lesion detection tasks.³³

Phantom Description

In this study, a 20 × 30 cm anthropomorphic phantom with extension rings that can bring the size to 25 × 35 cm and 30 × 40 cm (QRM, Moehrendorf, Germany) was used to represent the abdomen of small, medium, and large adult patients, respectively (Fig. 1). A 10-cm diameter water-equivalent phantom equipped with 4-mm and 8-mm diameter cylindrical holes was placed in the center of the abdominal phantom. The cylindrical holes were filled with iodinated contrast material (iohexol, Omnipaque 350; GE Healthcare Ireland, Cork, Ireland) to mimic low-contrast enhancing liver lesions that are commonly seen in patients.³⁴ The iodine solution was diluted to yield 4 concentrations (0.2, 0.5, 1.0, and 2.0 mg I/mL) to simulate lesions of various enhancement levels (CT number difference from the tissue-equivalent background of 4.5, 11.1, 22.0, and 44.2 HU at 120 kV, respectively). The 4 iodine concentrations were based on the reported thresholds, as low as 0.3 mg I/mL, for distinguishing clinically relevant contrast-enhanced pathologies using DECT technology.^{35–37} For the lesion detectability study, signal-absent images (without lesion) were generated by filling the holes with deionized water.

Image Acquisition and Data Preparation

A total of 15 unique conditions were scanned, including 4 iodine concentrations and a water scan (ie, zero iodine concentration) and 3 phantom sizes. Phantoms were scanned on a whole-body research PCD-CT scanner (SOMATOM CounT; Siemens Healthcare, Forchheim, Germany) and a second-generation dual-source CT scanner (SOMATOM Flash; Siemens Healthcare), which is the same scanner platform upon which the PCD-CT was built. Detailed descriptions of the PCD-CT system are given elsewhere.^{19,24} The tube potentials for both the SE and DE scans on the clinical EID-CT were selected based on our standard contrast-enhanced abdominal CT protocol (Table 1). Specifically, the SECT tube potential was determined by an automatic tube potential selection system (CARE kV;

Siemens Healthcare) with 120 reference tube potential and a strength setting of 8. CARE kV selects tube potential based on patient size and the strength setting, which is selected based on clinical task. The goal is to achieve the highest CNR and dose efficiency, with constrain that image noise cannot exceed a certain limit. For clinical tasks with high concentration of iodine, for example, CT angiography, higher strength setting (eg, 11) is used, which allows the scanner to select tube potential with highest CNR with little constrain on image noise. On the other hand, in noncontrast scan, lower strength setting (eg, 2) is selected where the noise constrain plays a more important role. Contrast-enhanced abdominal CT examinations are in between, with strength of 8 commonly used. The tube potential pairs used for DECT (80/Sn140 or 100/Sn140) were determined according to phantom size: 80/Sn140 for phantoms with lateral width of 33 cm or less and 100/Sn140 for phantoms with lateral width greater 33 cm. For PCD-CT, 140 kV and 25, 75 keV energy thresholds were used to scan all phantom sizes. Radiation dose was matched to that of routine abdominal CT examinations and was the same for PCD-CT and the commercial EID-CT scanner. To determine the dose levels, each phantom was scanned on the EID-CT system with a clinical protocol of 120 kV as reference kV (CARE kV; Siemens Healthcare) and 200 quality reference mAs using an automatic exposure control system (CARE Dose4D; Siemens Healthcare). The volume CT dose index ($CTDI_{vol}$) was 6.9, 10.8, and 17.7 mGy, and dose length product was 69, 108, and 177 mGy·cm for the small, medium, and large phantoms, respectively. For each phantom size, the tube current for the EID-CT and PCD-CT scans was adjusted to achieve the same $CTDI_{vol}$ (Table 1), thereby matching absorbed dose to the phantom across scanner types and acquisition protocols. The image reconstructions for both PCD-CT and EID-CT were matched and performed using a filtered back projection algorithm with a quantitative, medium smooth kernel (D30), 100-mm field of view, 512×512 matrix size, and 5-mm image thickness with 8-mm increments. For each condition, 15 repeated scans were performed, and 10 images were reconstructed per scan, yielding a total of 150 images per condition.

VMI Processing

The 2 energy-bin images from each PCD-CT acquisition and low- and high-energy images from each DE EID-CT acquisition were processed using research software (eXamine; Siemens Healthcare) to generate VMIs (Mono+; Siemens Healthcare) at energies of 50, 60, 70, and 80 keV. The VMIs at low keV range from Mono+ have been reported to have improved CNR by using a frequency-split processing technique.¹⁶ The optimal energy for DE EID-CT and PCD-CT was first determined using the CNR, and then determined using lesion detectability.

CNR Analysis

Circular regions of interest (ROIs) were drawn within the boundaries of the 8-mm diameter cylindrical holes, which contained either water or one of the iodine solutions, and in the tissue-equivalent background of the SE EID-CT images or the VMIs from the DE EID-CT and PCD-CT images. Contrast was measured as the CT number difference between the cylindrical holes and the tissue-equivalent background; noise was measured as the standard deviation of the CT numbers in the background region. Contrast-to-noise ratio values were

calculated and compared between SE EID-CT images, VMIs from DE EID-CT, and VMIs from PCD-CT for each condition.

Detectability Analysis

A CHO was used to assess the detectability of the contrast-enhanced lesions.^{33,38,39} The area under the receiver operator characteristic curve (AUC) was calculated using a Wilcoxon nonparametric method (see Text, Supplemental Digital Content 1, for the detailed descriptions of detectability analysis, <http://links.lww.com/RLI/A500>).

Human Observer Study

Although the performance of CHO has been shown to be correlated with that of human observer in previous studies, they focused on SE images acquired with EID-CT systems.^{33,38–40} Because this study includes PCD-CT acquisitions and postprocessed multienergy images, a 2AFC study with a subset of conditions was performed with 3 medical physicists to validate CHO results. Validation study with human observer was performed on a subset of data instead of the whole data set to limit the amount of effort. Each physicist is specialized in CT and having previously participated in such studies, serving as the human observers. Each observer was assigned to evaluate 6 conditions with the 4-mm lesion size (2 concentrations: 0.5 mg I/mL and 1.0 mg I/mL; 3 image types: SE, VMIs from EID-CT and PCD-CT). One hundred images were randomly selected from a total of 150 images for each condition, corresponding to 100 trials. Each trial consisted of an ROI with signal present at the center and an ROI with signal absent. The 2 alternative ROIs were presented side by side in a random order for observers to identify the one with the signal present. Observer studies were performed using a calibrated display and a customized user interface built with MATLAB (MathWorks, Natick, MA). The ratio between the number of correct selections and the total number of trials (100 in this study) was calculated as percent correct (PC) for each 2AFC task. The PC values for each condition were compared with the AUC values calculated in the CHO model observer studies under the same conditions, as the PC value in a 2AFC task is equal to the AUC in a detection task under Gaussian assumptions.

Statistical Analysis

Spearman correlation was applied to assess the agreement of performance between the CHO model observer and human observers. For each condition (4 iodine concentrations and 3 phantom sizes), 3 image types (SECT, optimal VMI from EID-CT, and optimal VMI from PCD-CT) were ranked from 1 to 3 based on their CNR values, with 1 being the highest and 3 being the lowest. The same ranking strategy was applied based on their AUC values for 4-mm and 8-mm lesion sizes, respectively. The average rankings for the 3 image types were compared using Friedman test independently to evaluate the overall performance difference for CNR and AUC rankings (combining 4-mm and 8-mm rankings), respectively. Conover post hoc testing with Bonferroni correction was used for pairwise comparisons when statistical significance was observed in Friedman test. All statistical analysis was performed using a free statistical package (R Project, Version 3.4.0; <http://www.r-project.org/>), with $P < 0.05$ considered to be statistically significant.

RESULTS

Selection for Optimal Monoenergetic Image

Figure 2 illustrated for the specific condition of a 4-mm lesion, 1.0 mg I/mL, and different phantom sizes, the corresponding CNR and AUC values of the PCD-CT VMIs were the highest at 50 keV. Similar trend was found for other conditions, and therefore, 50 keV was determined to be the optimal keV for VMIs in subsequent analyses.

Lesion Conspicuity Comparison

For the medium-sized (35 cm wide) phantom, representative PCD-CT VMIs at 50 keV, EID-CT VMIs at 50 keV, and SECT images for the 4-mm lesion and the evaluated iodine concentration levels are shown in Figure 3. For each image type, when compared with the water-only background images (0 mg/mL, left column in Fig. 3), the lesion conspicuity was improved with increasing iodine concentration. When compared across image types, both EID-CT and PCD-CT VMIs demonstrated superior iodine contrast but increased image noise compared with the corresponding SECT images.

Lesion CNR Comparison

Figure 4 demonstrated the CNR comparison across the different scanner/image types. Under the condition of 1.0 mg I/mL, 8-mm lesion size, and the medium-sized phantom, the CNR values of EID-CT VMIs (1.96 ± 0.52) and PCD-CT VMIs (2.01 ± 0.67) at 50 keV were both significantly higher ($P < 0.001$) than that of SECT images (1.11 ± 0.35). The CNR values for the 50 keV VMIs exceeded those for SECT for all other testing conditions.

Validation of CHO With Human Observer

For all 3 image types, with 0.5 mg I/mL and 1.0 mg I/mL concentrations in the 35 cm phantom, the correlation between the performance of CHO model observer and the human observers (Fig. 5) was significant ($\rho = 0.91$, $P < 0.01$), supporting the ability of the CHO model observer analysis of lesion detectability to estimate human observer performance in a 2AFC task.

Comparison of Lesion Detection Using Validated CHO

Figure 6 illustrated the comparison of AUC for detectability of the 4-mm lesions. At the condition of 1.0 mg I/mL, 4 mm lesion, and medium-sized phantom, PCD-CT VMIs at 50 keV had a comparable AUC (0.98 ± 0.01) to that of SECT images (0.98 ± 0.01 , $P > 0.05$) and both were slightly lower than DE VMI at 50 keV (0.99 ± 0.01 , $P < 0.01$). For the 8-mm lesion, all 3 image types demonstrated saturated AUC (1.00 ± 0.00) for 1.0 mg I/mL and the medium-sized phantom, indicating a perfect detection of lesion at this condition (Fig. 7). For other conditions (iodine concentrations and phantom sizes), results of the AUC comparison among the 3 image types followed a similar trend except 0.2 mg/cc concentration.

Overall Performance Comparison Among EID-CT and PCD-CT Protocols

The Friedman test showed statistically significant differences ($P < 0.0001$) in CNR rankings among the 3 image types (Table 2). In the pairwise comparisons, VMIs at 50 keV from both

DE (average ranking = 1.6 ± 0.5) and PCD (average ranking = 1.3 ± 0.5) demonstrated a significant higher ($P < 0.0001$) ranking of CNR than SECT images (average ranking = 3.0 ± 0.0). There was no statistical difference ($P > 0.05$) in CNR rankings between PCD-CT VMIs and EID-CT VMIs at 50 keV. When overall AUC rankings (combining for 4-mm and 8-mm lesions) were compared across 3 image types (average ranking: EID-CT SE = 1.5 ± 0.8 ; EID-CT VMI = 1.3 ± 0.4 ; PCD-CT VMI = 1.6 ± 0.8), Friedman test showed no statistical difference ($P > 0.05$), indicating that the overall detectability performance of the 3 image types was comparable.

DISCUSSION

We investigated the feasibility of a single universal acquisition and reconstruction protocol (140 kV, 25 keV and 75 keV thresholds, and 50 keV VMI) for a range of conditions mimicking CT scans of adults for the task of low contrast liver lesion detection on a PCD-CT to simplify the clinical workflow. Our results demonstrated that, using the proposed universal PCD-CT protocol or the patient size-dependent EID-CT, iodine CNR and detectability of VMIs were either greater than or comparable to those of SECT images acquired at the optimal tube potential; this was true for all 3 patient sizes.

In current practice, tube potential selection in CT examinations is based on knowledge of patient size, such as weight, body mass index, lateral size, or attenuation information from a CT localizer radiograph. Several phantom and clinical studies have demonstrated the dose reduction and other clinical benefits that can be achieved with the use of lower tube potentials for abdominal CT examinations.^{2,4,41} However, if a lower tube potential is inappropriately selected, especially for large patients, this could lead to suboptimal image quality and/or degraded detection and characterization of pathology.⁵ Attenuation-based automated tube potential selection has been successfully implanted in several clinical studies, and it provides a task-driven approach to increasing iodine contrast while maintaining acceptable image noise.^{42,43} However, this function is only available from a few scanner manufacturers and, even then, only for designated scanner models. In addition, factors such as the heterogeneity of patient attenuation along the cranial-caudal direction and tube power limits can make the task difficult to perform in several different scenarios. Although it is small, the risk of using inappropriate tube potential settings still exists with automated tube potential selection. The method proposed in this study utilized the most penetrating tube potential (140 kV) available on the evaluated system for all patient sizes to avoid any chances for suboptimal image quality (eg, image artifacts and excessive noise) due to inappropriate selection of a tube potential that is too low for patients who are large in an overall sense or for patients that are large at key anatomic levels (eg, patients who carry additional weight primarily at their hips, but who are otherwise of average [medium] size). The high tube potential (140 kV) also provides better spectral separation between different energy bins, which consequently results in better image quality in VMIs. The tube current on the PCD-CT with 140 kV were adjusted to match the clinical abdominal dose; therefore, the proposed method is dose neutral. In addition, the collection of multienergy data using PCD-CT or DE approaches enables the generation of low-energy VMIs (eg, 50 keV), which is close to the iodine k-edge and therefore boosts the iodine signal, resulting in comparable or higher iodine CNR and detectability compared with SECT with an optimal tube potential.

Due to its superior performance in terms of iodine CNR and low-contrast lesion detectability, 50 keV was selected as the optimal energy for a universal PCD-CT protocol. The iodine signal in VMIs increases with decreased photon energy; however, the image noise also increases at lower VMI energy settings.⁴⁴ To address the issue of increased noise, energy domain noise reduction techniques in DECT have been used in VMIs.^{16,44,45} It has been shown that an optimal VMI energy exists that has the highest iodine CNR.^{46,47} Our results showed that PCD-CT VMIs at 50 keV had at least 15% greater iodine CNR than those at higher VMI energy levels. Meanwhile, the iodine detectability for PCD-CT VMIs at 50 keV was equal to or slightly higher than higher energy levels; 40 keV was excluded in the analysis because of its excessive image noise and the presence of artifacts in the large phantoms.⁴⁶

When compared with conventional SECT images with size-specific tube potential selection, PCD-CT 50 keV VMIs demonstrated superior CNR. Due to the known discrepancy between iodine CNR and reader performance for low-contrast lesion detection,³⁰ we further tested iodine detectability using a previously validated model observer. In addition, validation data were collected for this specific study to confirm that the use of a CHO model observer provides data that can be used to guide these protocol optimization questions when a PCD-CT system is involved. Results suggested that lesion detectability was slightly improved for PCD-CT VMIs when compared with conventional SECT images across 3 adult abdomen sizes. In general, the difference in detectability among scanner models was less than that shown in the CNR comparison, emphasizing the importance of evaluating detectability in such studies due to the limitations of CNR in nonlinear algorithms.³⁰ These findings suggest that all adult patients, regardless of size, can be scanned on the PCD-CT using a universal acquisition protocol at the tube potential of 140 kV and reconstructed with 50 keV VMI. The proposed approach will significantly simplify the clinical workflow and reduce inadvertent selection of an inferior kV setting.

Compared with EID-DECT, prior work using simulations predicted slightly lower spectral performance for PCD-CT, with 5% higher noise in the iodine quantification maps.⁴⁸ A previous phantom study showed that the spectral performance was comparable between EID-DECT and PCD-CT in terms of the iodine quantification accuracy and VMI CT number accuracy.⁴⁹ In this study, we further tested that iodine CNR and detectability were similar between VMIs generated from size-specific EID-CT protocols and the universal PCD-CT protocol across different phantom sizes.

There are some limitations of this study. First, only phantom experiments with cylindrical lesions were investigated. Real lesions may have different shapes, for example, sphere or irregular shape. Second, the PCD-CT system was not equipped with dose modulation during our phantom investigations. Future patient studies with dose modulation function (once available), inclusion of different lesion shapes, and more complex anatomical background are underway to confirm the clinical performance of the proposed universal PCD protocol across different patient sizes. Third, this study focused on contrast-enhanced abdominal CT examinations where 50 keV VMIs were determined to appropriately balance iodine contrast and image noise. For other examination types, such as CT angiography, the optimal VMI energy for all patient sizes might be different, even though the acquisition protocol (140 kV,

25 and 75 keV energy thresholds) is the same. However, because detection of liver lesions is one of the most noise-sensitive examinations, we suspect that examinations used for higher-contrast tasks stand to benefit even more at 50 keV, because the increase in image noise will be much less of a concern. Nonetheless, further investigation will be required to determine the use of a universal protocol and the optimal VMI energy for other types of examinations. Fourth, this study focused on adult patients with abdominal sizes that are commonly seen in clinical practice. For SECT, the automatic tube potential selection software picked 100 kV for all phantom sizes, which is a potential limitation of this study. Further validation will be necessary for those uncommonly seen extreme large or small patient sizes that the software may pick different tube potentials. Finally, the concept of a universal PCD-CT protocol is equally compelling for the pediatric patient population, although the optimal tube potential, energy thresholds, and VMI energy might not be the same as determined here for adult patients. Investigations using phantoms that mimic pediatric sizes are also underway.

CONCLUSIONS

Fifty-kiloelectrovolt VMIs on a whole-body PCD-CT system acquired with a universal protocol demonstrated iodine CNR and detectability greater than or comparable to those of traditional SECT images and EID-DECT VMIs across adult patient sizes. Although SECT and EID-DECT examinations require the use of patient size-specific acquisition settings, which can complicate workflows and/or lead to degraded image quality, our results suggest that, using PCD-CT technology, a much simpler approach to simplifying protocol selection can be adopted, namely, the use of a single tube potential (140 kV), fixed energy thresholds (25 and 75 keV), and a VMI energy setting of 50 keV results in iodine CNR and low-contrast lesion detectability as good as or better than that patient size-specific protocols across adult patient sizes.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

ACKNOWLEDGMENTS

The authors would like to thank Ms Kristina Nunez and Lucy Bahn, PhD, for their help with manuscript preparation.

Conflicts of interest and sources of funding: The project described was supported by the National Institutes of Health under award numbers R01 EB016966 and C06 RR018898, and in collaboration with Siemens Healthcare. The content is solely the responsibility of the authors and does not necessarily represent the official views of the National Institutes of Health. The device described is a research scanner and not commercially available. Dr McCollough receives industry funding from Siemens Healthcare. For the remaining authors, none were declared.

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FIGURE 1.

The anthropomorphic abdominal phantom and extension rings used to emulate small (lateral width = 30 cm), medium (lateral width = 35 cm), and large (lateral width = 40 cm) adults. The 4-mm and 8-mm diameter cylindrical holes in the central 10-cm insert were filled with iodine solutions to mimic contrast-enhanced lesions.

FIGURE 2.

Iodine contrast-to-noise ratio (CNR) and detectability (AUC) of the 4-mm, 1.0 mg I/mL lesion measured from photon-counting detector (PCD) virtual monoenergetic images (VMIs) at various energy levels in small (30-cm wide), medium (35-cm wide), and large (40-cm wide) phantom sizes.

FIGURE 3.

Representative images of a 4-mm lesion with various iodine concentrations (0, 0.2, 0.5, 1.0, 2.0 mg I/mL) and a 35-cm wide phantom. PCD indicates photon-counting detector; EID, energy-integrating detector; DE, dual-energy; SE, single-energy; VMI, virtual monoenergetic image. WW/WL = 300/40 HU.

FIGURE 4.

Iodine contrast-to-noise ratio (CNR) comparisons between photon-counting detector (PCD) CT virtual monoenergetic images (VMIs) at 50 keV, energy-integrating detector (EID) dual-energy (DE) VMI and single-energy (SE) images across different phantom sizes (30, 35, 40 cm) and iodine concentrations (0.2, 0.5, 1.0, 2.0 mg I/mL).

FIGURE 5.

Comparison between human observer detection performance (solid triangles) with predicted performance by a channelized Hotelling observer (CHO, empty squares) for 6 selected conditions with a 4-mm lesion size: 2 concentrations (0.5 mgI/mL and 1.0 mgI/mL) and 3 image types. PCD indicates photon-counting detector; EID, energy-integrating detector; DE, dual-energy; SE, single-energy; VMI, virtual monoenergetic image.

FIGURE 6.

Area under the curve (AUC) for detectability of the 4-mm lesions calculated from channelized Hotelling observer model across different phantom sizes (30, 35, 40 cm) and iodine concentrations (0.2, 0.5, 1.0, 2.0 mg I/mL). PCD indicates photon-counting detector; EID, energy-integrating detector; DE, dual-energy; SE, single-energy; VMI, virtual monoenergetic image.

FIGURE 7.

Area under the curve (AUC) for detectability of the 8-mm lesions calculated from channelized Hotelling observer model across different phantom sizes (30, 35, 40 cm) and iodine concentrations (0.2, 0.5, 1.0, 2.0 mg I/mL). PCD indicates photon-counting detector; EID, energy-integrating detector; DE, dual-energy; SE, single-energy; VMI, virtual monoenergetic image.

TABLE 1.
Acquisition Protocols and Radiation Doses for PCD and EID-DE, SE Scans Performed With Variable Adult Sizes

Patient Lateral Size	PCD	EID-DE	EID-SE	CTDI _{vol} , mGy	DLP, mGy·cm
30 cm	140 kV (25, 75 keV)	80/140Sn kV	100 kV	6.9	69
35 cm		100/140Sn kV	100 kV	10.8	108
40 cm		100/140Sn kV	100 kV	17.7	177

PCD indicates photon-counting detector; EID, energy-integrating detector; DE, dual-energy; SE, single-energy; CTDI_{vol}, CT dose index; DLP, dose length product.

TABLE 2.
Comparison of Average Ranks for CNR and AUC Among PCD VMI at 50 keV, DE VMI at 50 keV, and SE Images

Average Rank				Pairwise Comparison <i>P</i>			
	PCD VMI	DE VMI	SE	Friedman Test <i>P</i>	PCD VMI vs SE	DE VMI vs SE	PCD VMI vs DE VMI
CNR	1.3	1.6	3.0	<0.0001	<0.0001	<0.0001	0.25
AUC	1.6	1.3	1.5	0.19	NA	NA	NA

Pairwise comparison was performed when Friedman test was statistically significant. $P < 0.05$ was considered as statistically significant. CNR, contrast-to-noise ratio; AUC, area under the receiver operating characteristic curve; PCD, photon-counting detector; VMI, virtual monoenergetic image; DE, dual-energy; SE, single-energy; NA, not available.