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Initial results from a prototype whole-body photon-counting computed tomography system

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Abstract

X-ray computed tomography (CT) with energy-discriminating capabilities presents exciting opportunities for increased dose efficiency and improved material decomposition analyses. However, due to constraints imposed by the inability of photon-counting detectors (PCD) to respond accurately at high photon flux, to date there has been no clinical application of PCD-CT. Recently, our lab installed a research prototype system consisting of two x-ray sources and two corresponding detectors, one using an energy-integrating detector (EID) and the other using a PCD. In this work, we report the first third-party evaluation of this prototype CT system using both phantoms and a cadaver head. The phantom studies demonstrated several promising characteristics of the PCD sub-system, including improved longitudinal spatial resolution and reduced beam hardening artifacts, relative to the EID sub-system. More importantly, we found that the PCD sub-system offers excellent pulse pileup control in cases of x-ray flux up to 550 mA at 140 kV, which corresponds to approximately 2.5×10^{11} photons per cm² per second. In an anthropomorphic phantom and a cadaver head, the PCD sub-system provided image quality comparable to the EID sub-system for the same dose level. Our results demonstrate the potential of the prototype system to produce clinically-acceptable images *in vivo*.

Keywords

Photon-Counting CT; Research Prototype; System Evaluation

1. PURPOSE

X-ray computed tomography (CT) using photon-counting detector (PCD) technology has been investigated by a number of research teams [1–3]. Compared to an energy-integrating detector (EID), PCDs have the advantage of significantly reduced electronic noise and increased contrast-to-noise ratio (CNR); they can also provide information about the energy of detected photons. These advantages present exciting opportunities for increased radiation dose efficiency and improved material decomposition analyses, and may also provide opportunities for new clinical applications. However, the performance of current PCD technology is degraded due to factors such as charge sharing, pulse pileup, and K-escape [4]. Some of these factors, particularly pulse pileup, are more pronounced in cases of high x-ray

flux, which has presented a significant barrier to the development of a clinically-viable PCD CT system.

Recently, a whole-body PCD research prototype CT system was installed and commissioned in our laboratory [5]. This system was developed using a dual-source CT system [6]. Both x-ray sources use the same type of x-ray tube, but are associated with different types of detectors: one source is coupled with an EID detector array, and the other is coupled with a PCD detector array. To date, studies performed on this system have been reported only by the system manufacturer. This work provides the first third-party evaluation of the research prototype system. It uses image quality phantoms, an anthropomorphic phantom, and a human cadaver head to measure and compare the performance between the commercial EID system and the prototype PCD system, at dose levels similar to those of realistic clinical scenarios.

2. MATERIALS AND METHODS

The whole-body prototype PCD-CT system consists of two sub-systems: the EID sub-system and the PCD sub-system. The EID system has a 500 mm field-of-view (FOV) and 38.4 mm longitudinal coverage (at iso-center along the direction of the patient table), whereas the PCD system has a 275 mm FOV and 14 mm longitudinal coverage. The EID and PCD systems have slightly different pixel sizes at the iso-center, with the former being 0.6 mm and the latter being 0.5 mm. The physical dimensions of the PCD pixels are 0.9 mm by 1.1 mm, each consisting of 4 by 4 square sub-pixels with an edge length of 0.225 mm. The 1.1 mm pixel pitch is due to the inclusion of an x-ray collimator between each pixel in the circumferential direction (the collimator width is approximately that of 1 sub-pixel). Each sub-pixel admits two thresholds, which enables the multi-energy data acquisition capability of the PCD. Currently, the PCD system provides two data acquisition modes [5], referred to as the macro mode, which counts number of photons above two energy thresholds (with all 16 sub-pixels using the same thresholds), and the chess mode, which counts number of photons above four energy thresholds (with 8 sub-pixels using two thresholds and the other 8 using another two thresholds). In the macro mode, two threshold images and two energy bin images are produced; in the chess mode, four threshold images and four energy bin images are produced. In each mode, the image produced using the highest threshold settings is equivalent to the image of the highest energy bin. Further details regarding the prototype PCD-CT system can be found in [7].

The system was evaluated by considering the following assessments: CT number accuracy, CT number uniformity, spatial resolution, impact of pulse pileup, and clinical feasibility. All measurements were performed in an almost identical manner between the EID and PCD systems, except that the former used 32×0.6 mm collimation and the latter used 32×0.5 mm collimation. Note that, the first and last two rows are discarded in the PCD, resulting in 14 mm coverage along the longitudinal direction.

The CT numbers of water, bone, and iodine were measured using three phantoms: a cylindrical water phantom, the ACR CT accreditation phantom (model 464, Gammex-RMI, Middleton, WI), and a torso-shaped water phantom with iodine solution insert (Figure 1).

All three phantoms were scanned at 210 mAs and 140 kV on both the EID and PCD systems. The macro mode with thresholds [25, 65] keV was used on the PCD system. All images were reconstructed using an image thickness of 1 mm and a medium smooth body kernel (D30). For the cylindrical water tank with 200 mm diameter, CT numbers were measured from a central region-of-interest (ROI) and four peripheral ROIs to assess uniformity, as shown in Figure 1b. For the ACR CT accreditation phantom, the CT number of the bone-mimicking material was measured. For the torso-shaped water tank that was 200 mm wide in the lateral direction, a vial with iodine solution of 25 mg/mL concentration was inserted into its center. CT numbers were measured from the iodine solution.

The in-plane spatial resolution was measured in terms of the modulation transfer function (MTF), whereas the longitudinal spatial resolution was measured in terms of the slice sensitivity profile (SSP). Both measurements were achieved using the spiral mode with a pitch of 0.5. For the PCD system, only the macro mode acquisition was used, with energy thresholds of [25, 65] keV. The MTF was measured using a 0.125 mm diameter tantalum wire (Figure 2 *left*) that was scanned using 40 mAs and 140 kV. The 1 mm images were reconstructed with a 50 mm FOV. The SSP was measured using a 0.05 mm thick gold foil (QRM GmbH, Moehrendorf, Germany; see Figure 2 *right*) that was scanned at 70 mAs and 140 kV. Both EID and PCD images were reconstructed with a 50 mm FOV and an image increment of 0.1 mm. The thinnest possible slice thickness was used for image reconstruction for each system, which was 0.6 mm for the EID system and 0.5 mm for the PCD system (slightly varied due to different collimations).

To assess the impact of pulse pileup at high photon flux, we measured the change in mean CT number and image noise (standard deviation of CT numbers in an ROI). A small torso-shaped water phantom (Lateral width: 7.7 cm, Figure 3 *left*) was scanned at 140 kV; the PCD data were acquired in the macro mode with energy thresholds at [25, 65] keV. Data were acquired from 20 mA to 548 mA at intervals of 12 mA. At each mA setting, the scan was repeated, such that two sets of images were reconstructed. These images were subtracted to form a set of difference images from which the noise within an ROI was measured. The ratio of the noise in bin 1 ([25, 65] keV) to the noise in bin 2 ([65, 140] keV) was then calculated. In the absence of pulse pileup effects and with a fixed kV and linearly increasing mA, image noise would be expected to decrease following an inverse square root function. Further, in the absence of pileup effects, the ratio of the noise in bin 1 to the noise in bin 2 would be constant. At high photon flux, severe pulse pileup effects, such as two bin-1 photons being counted as a single bin-2 photon, would cause noticeable deviations from these expected results.

Finally, to assess clinical feasibility, an anthropomorphic thorax phantom (model 007TE, CIRS, Norfolk, Virginia; see Figure 3 *right*) and a cadaver head were scanned on both the EID and PCD systems. The CIRS phantom represents the size of a 15 year old adult (Anterior-posterior height: 18.4 cm, Lateral width: 23.9 cm) and could be fully contained in the FOV of the PCD sub-system. Two inserts were added to the CIRS phantom: bone (200 mg/cc) and iodine (20 mg/mL). The phantom was scanned at 46 mAs and 140 kV, with the CTDI_{vol} matching our clinical protocol for this size patient. Images of the CIRS phantom were reconstructed using a 5 mm image thickness and a medium smooth body kernel (D30).

Data were acquired in the macro mode with energy thresholds at [25, 65] keV. The cadaver head was scanned at 550 mAs and 140 kV. The PCD data were acquired in the macro mode with energy thresholds at [20, 63] keV.

3. RESULTS

In this section, results regarding CT number and spatial resolution will be presented first. Then, the impact of pulse pileup will be reported. Finally, clinical feasibility of the PCD system will be demonstrated by images of the CIRS phantom and the cadaver head.

3.1 CT number accuracy and uniformity

The CT number accuracy and uniformity in the water phantom were clinically acceptable for the PCD system; see Figure 4 **left and middle**. In terms of CT number uniformity, the EID system was slightly better than the PCD system, but both were within the clinically acceptable range (± 5 HU). The difference in CT numbers between the EID and PCD systems for high-Z materials, such as bone and iodine, was consistent with expectations; see Figure 4 *right*. For example, the CT number of iodine in the PCD threshold-low ([25, 140] keV) images was greater than that in the EID images. This is because the projection data from a PCD system are in terms of photon-counts, whereas those from an EID system are energy weighted photon counts. Therefore, the PCD x-ray spectrum is weighted more heavily in the low energy than the EID x-ray spectrum, resulting in greater CT numbers for high-Z materials [8]. With accurate CT numbers for water and increased CT numbers for iodine, the PCD images presented higher iodine contrast than the EID images. Moreover, the PCD images in different energy ranges presented different CT number for high-Z materials. For example, the CT number of bone in bin 1 ([25, 65] keV) was greater than that in bin 2 ([65, 140] keV), which is strong evidence that the PCD system provided energy-selective information.

3.2 Spatial resolution

The measured in-plane MTF curves and longitudinal SSP are shown in Figure 5. The EID and PCD systems appear to have similar in-plane resolution, which is consistent with the fact that the vendor has purposely matched the in-plane resolution when the same reconstruction kernel is used for both systems. With the same reconstruction kernel, the best-achievable SSP was better on the PCD system than on the EID system: the full widths at half maximum (FWHM) of the line spread function along the longitudinal direction were 0.850 mm and 0.625 mm for the EID and PCD systems, respectively. The resultant better SSP is mainly because a smaller detector pixel size was used in the PCD system.

3.3 Pulse pileup

The CT numbers of water from the first set of scans (i.e. unsubtracted), image noise from the two bins, the fitted noise-to-mA curves for bin 1 and bin 2, and the ratios of the noise of bin 1 to bin 2 are shown in Figure 6 (left, middle, right, respectively). Several observations can be made from Figure 6. First, the CT numbers of water of both bin 1 and bin 2 images start to increase when mA increases above 300, and the value in bin 1 increased faster than in bin 2. This is consistent with the observation from the middle plot of Figure 6 that, at 140 kV

and with a very small phantom, when mA increases above 300, the noise of bin-2 started to deviate slightly above the fitted curve, whereas the noise of bin 1 started to deviate upwards from the fitted curve. This phenomenon is due to pulse pileup causing, for example, two bin-1 photons to be counted as a single bin-2 photon. These observations indicate that the impact of pulse pileup becomes noticeable only above 300 mA, again for 140 kV and a very small phantom. Second, the noise of bin 1 is slightly smaller than bin 2, which might be partially due to the fact that bin 1 had slightly more counts than bin 2. Third, the ratio of the noise of bin 1 to bin 2 is a good indicator of dose distribution and dose efficiency. In Figure 6 right, the ratio tends to increase when mA increases, which is consistent with pileup shifting the pulse spectrum from bin 1 toward bin 2. Note that the deviation of the noise from fitted inverse square root function was small, demonstrating that pulse pileup had a reasonably low impact.

3.4 Clinical feasibility

For the CIRS phantom, we observed similar image quality between the EID images and the PCD threshold-low (all photons from the low threshold and higher) images; see the left and middle in Figure 7. As expected, bin 2 images showed significantly fewer beam hardening artifacts than the other images due to being restricted to only the higher energy photons. The price paid is increased noise, since bin 2 images only used a portion of the total detected photons. Similar observations can be made for images of the cadaver head; see Figure 8. Both the CIRS and cadaver head images showed promising overall performance of the PCD system in a realistic clinical setting.

4. CONCLUSIONS

We have presented the initial results of the first third-party evaluation of a whole-body PCCT prototype system. For water, the PCD system in macro mode provided CT number accuracy (± 2.5 HU) and uniformity (± 4 HU) that met clinical requirements. For high-Z materials, the PCD system delivered energy-selective information and presented improved high contrast when compared to the EID system. In terms of spatial resolution, the PCD system and the EID system had similar in-plane resolution, but the former excelled the latter in the best-achievable SSP in the longitudinal direction. Very importantly, we have shown that the PCD system suffered from negligible pileup effects in extreme cases, i.e., 7.7 cm wide water phantom at high x-ray flux up to 550 mA at 140 kV. The PCD threshold-low images of the anthropomorphic thorax phantom and the cadaver head showed similar image quality to the EID images. Moreover, the PCD bin images delivered energy-selective information that provided additional benefits, such as reduction of beam hardening artifacts. Overall, the results presented in this paper are strong evidence that the whole-body research PCCT prototype system can provide clinically acceptable image quality, even at clinically realistic and extremely high x-ray flux values.

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The equipment and concepts described in this work are based on a research prototype and are not commercially available.

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Figure 1.

Phantoms used for CT number measurement. a): cylindrical tank of 200 mm diameter filled with water. b) illustration for uniformity measurement. c): the ACR CT accreditation phantom. d): a tube filled with iodine concentration placed at the center of a water tank. The iodine concentration was 25 mg/mL; the water tank was 200 mm wide in the lateral direction.

Figure 2. Phantoms used for spatial resolution measurement. Left: a 0.125 tantalum wire for measurement of in-plane modulation transfer function. Right: a 0.05 mm gold foil for measurement of longitudinal slice sensitivity profile.

Figure 3.

Left: A small torso-shaped phantom filled with water (lateral width.: 7.7 cm). Right: A CIRS thorax phantom mimicking a 15 year old human being (Anterior-posterior height: 18.4 cm, Lateral width: 23.9 cm).

Figure 4. Left to right: water CT number accuracy, water CT number uniformity, and CT number accuracy for bone and iodine.

Figure 5. Measurements of spatial resolution. Left: modulation transfer function (MTF). Right: slice sensitivity profile (SSP).

Figure 6.

Left: CT number versus mA; middle: noise (std) versus mA; right: ratio of the noise of bin 1 to that of bin 2 versus mA.

Figure 7.
Reconstructions of the CIRS phantom. Display window: $[-100, 100]$ HU.

Figure 8.
Reconstructed images of two regions within the cadaver head. Left: EID; middle: [20, 140] keV; right: [63, 140] keV. Display window: [-100, 200] HU. Note the reduced beam hardening artifacts in the high-energy PCD bin image.