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Ultra-High Resolution Photon-Counting Detector CT Reconstruction using Spectral Prior Image Constrained Compressed-Sensing (UHR-SPICCS)

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Abstract

Photon-counting detector based CT (PCD-CT) enables dose efficient high resolution imaging, in addition to providing multi-energy information. This allows better delineation of anatomical structures crucial for several clinical applications ranging from temporal bone imaging to pulmonary nodule visualization. Due to the smaller detector pixel sizes required for high resolution imaging, the PCD-CT images suffer from higher noise levels. The image quality is further degraded in narrow energy bins as a consequence of low photon counts. This limits the potential benefits that high-resolution PCD-CT could offer. Conventional reconstruction techniques such as the filtered back projection (FBP) have poor performance when reconstructing noisy CT projection data. To enable low noise multi-energy reconstructions, we employed a spectral prior image constrained compressed sensing (SPICCS) framework that exploits the spatio-spectral redundancy in the multi-energy acquisitions. We demonstrated noise reduction in narrow energy bins without losing energy-specific attenuation information and spatial resolution. We scanned an anthropomorphic head phantom, and a euthanized pig using our whole-body prototype PCD-CT system in the ultra-high resolution mode at 120kV. Image reconstructions were performed using SPICCS and compared with conventional FBP. Noise reduction of 18 to 46% was noticed in narrow energy bins corresponding to 25 – 65 keV and 65 – 120 keV, while the mean CT number was preserved. Spatial resolution measurement showed similar modulation transfer function (MTF) values between FBP and SPICCS, demonstrating preservation of spatial resolution.

Keywords

Photon-counting detectors; computed tomography; multi-energy; ultra-high resolution; image reconstruction; compressed sensing; prior image constrained compressed sensing

1. INTRODUCTION

Photon counting detector-based CT (PCD-CT) enables energy-resolved, multi-energy acquisitions using pixelated direct conversion detectors. Simultaneous multi-energy CT

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projection data are acquired from a single exposure, subsequently allowing the formation of basis material images, virtual non-contrast images, and virtual mono-energetic images. In addition, photon-counting detectors do not suffer from the fill factor limitations inherent to the traditional energy-integration detectors, thereby allowing the design of smaller detector pixels to achieve dose-efficient high resolution CT without the need for comb/grid filters. Recently Leng *et al.*^{1,2} demonstrated the high resolution acquisition modes of a whole-body research PCD-CT system. Due to smaller per-pixel area in combination with narrow energy bin required for higher image contrast and better spectral sampling, the high resolution modes are inherently prone to higher noise levels in the energy bin images. Conventional reconstruction techniques such as filtered back projection (FBP) have limited performance when processing noisy PCD-CT data. Multi-energy data from photon-counting detectors (PCD) exhibit high spatio-spectral correlation, and exploiting this data redundancy could yield low noise PCD-CT reconstructions^{3,4}. We describe a spectral prior image-constrained compressed sensing (UHR-SPICCS) framework⁴ to enable low noise, high resolution CT reconstruction from a whole-body PCD-CT system.

1.1 Whole-body photon-counting CT system

The research whole-body PCD-CT^{5,6} is a modified second-generation dual-source system (Somatom Definition Flash, Siemens Healthcare, Forchheim, Germany). One of the two subsystems is equipped with a conventional energy-integrating detector (EID) while the other subsystem incorporates a cadmium telluride photon-counting detector array. The in-plane field of view (FOV) is 27.5 cm for the PCD subsystem at scan isocenter, and 50 cm for the EID subsystem. For scanning objects with diameters larger than the PCD FOV, an additional low dose data completion scan (DCS) using the EID system can be performed for correcting the data truncation⁷.

Four CT acquisition modes are available in the PCD-CT system: (i) Macro mode with two energy thresholds for dual-energy acquisitions at a detector pixel size of 0.5 mm at scan isocenter (ii) Chess mode with four energy thresholds for multi-energy acquisitions at a detector pixel size of 0.5 mm at scan isocenter (iii) Ultra-high resolution (UHR) mode with two energy thresholds at a detector pixel size of 0.25 mm at scan isocenter. The x-ray focal spot size for this mode is reduced to 0.7 mm from the standard 0.9 mm to minimize focal blurring required to achieve high resolution imaging, and (iv) Sharp mode with two energy thresholds at a detector pixel size of 0.25 mm at scan isocenter for the low energy threshold (< 20 keV, < 50 keV) and 0.5 mm pixel size at scan isocenter for high energy threshold (> 50 keV, > 90 keV). In this work, we focused on denoising the UHR mode data which provides the highest spatial resolution but also suffer from higher noise levels compared to the other PCD-CT acquisition modes.

2. MATERIALS AND METHODS

2.1 Spectral Prior Image Constrained Compressed Sensing (SPICCS)

The SPICCS^{4,8} reconstruction framework reconstructs the energy-selective (i.e. energy bin) images by exploiting the spatio-spectral redundancy across the acquired multi-energy

projection data. It attempts to find the optimal energy bin image ($\underline{x}^*$) by solving the following constrained minimization problem:

$$\underline{x}^* = \arg \min f_c(\underline{x}) \text{ s.t. } A\underline{x} = \underline{b} \quad (1)$$

where the objective function is defined as:

$$f_c(\underline{x}) = c \|\underline{x}\|_{TV} + (1 - c) \|\underline{x} - \underline{x}_{prior}\|_{TV} \quad (2)$$

Here, $\underline{x}$ is the image vector representing the linear attenuation coefficient distribution within an energy bin of interest; $\underline{b}$ is a column vector containing the acquired projection data for the corresponding energy bin; $\underline{x}_{prior}$ denotes the prior image which can be obtained from the acquired full spectrum PCD data, and therefore possesses minimal noise compared to the individual energy bin images. TV stands for total variance and c is a control parameter/weighting factor ($0 < c < 1$).

Eq. 1 is solved via a two-step alternating iteration routine. Firstly, data consistency ($A\underline{x} = \underline{b}$) and image non-negativity are enforced using algebraic reconstruction technique and convexity projection. Secondly, the objective function from Eq. 2 is minimized using a gradient descent algorithm. The stopping criterion is set such that the difference images reconstructed between two consecutive iterations, normalized to the prior image, should be below a predefined threshold (T). This is given by

$$R = \frac{\|I_{SART}^k - I_{SART}^{k-1}\|}{\|I_{FBP}^{Prior}\|}$$

Here I is the 3D image volume, k is the iteration number, $\|\cdot\|$ is the L_2 norm, and convergence is defined by $R < T$, where $T \ll 1$.

A value of $c = 0.5$ (Eq. 2) and $T = 0.005$ was empirically determined from a previous study⁴. The number of iterations for the TV subroutine was fixed at 50. The full spectrum prior image volume was generated using conventional FBP technique using a ramp filter apodized with a Hanning function. The SPICCS reconstructions were performed in 3D cone-beam geometry with axial scans.

2.2 Experimental studies

PCD-CT of anthropomorphic head phantom—An anthropomorphic head phantom was scanned on the PCD-CT system using the UHR mode. The phantom diameter was smaller than the in-plane FOV of the PCD system; therefore a DCS was not required. Axial acquisitions were performed at 120 kV, 132 mAs (CTDIvol(16cm) = 9.5 mGy) and 1s rotation time. The energy thresholds were set to be 25 and 65 keV. Spectral preservation was assessed in the bin images by comparing the mean CT number in a representative regions-

of-interest (ROI) from conventional FBP, and UHR-SPICCS. Similarly, noise reduction was quantified using standard deviation of CT numbers measured in an ROI between FBP and UHR-SPICCS.

Spatial resolution—To demonstrate preservation of spatial resolution in UHR-SPICCS reconstructions, we measured the modulation transfer function (MTF) in the FBP and UHR-SPICCS reconstructed images. For this purpose, a thin wire phantom made from tantalum (0.125 mm diameter) placed at the scan isocenter was scanned using the UHR mode at 120 kV. The wire contained in a vial was placed inside a 20cm solid CT water phantom. The energy bin image corresponding to 25 – 65 keV reconstructed at 1024×1024 matrix size and 0.25 mm slice thickness was used to assess the spatial resolution in the images produced by UHR-SPICCS and conventional FBP. The MTF measurement for the UHR mode was also compared to the results from Macro mode at standard resolution from a previous study⁴.

PCD-CT of pig—A freshly euthanized pig was scanned on the PCD-CT system using the UHR mode of the PCD-CT system. The head and neck region of the pig was scanned within the in-plane FOV of the PCD system. Axial acquisitions were performed using 120 kV, 148 mAs (CTDIvol(32cm) = 12 mGy) and 1s rotation time. The energy thresholds were set to be 25 and 65 keV. Similar to the anthropomorphic head phantom scan, spectral preservation and noise reduction from UHR-SPICCS were quantified in representative ROIs using mean CT number and standard deviation, respectively, and compared to conventional FBP images.

3. RESULTS AND DISCUSSION

The FBP and UHR-SPICCS reconstructed slices from the anthropomorphic head phantom scan are shown in Fig 1. The full-spectrum image (25 – 120 keV) from FBP was used as the prior, and narrow energy bins at energy ranges 25 – 65 keV and 65 – 120 keV were iteratively reconstructed using UHR-SPICCS. For comparison, FBP images corresponding to the narrow bins are also shown. Analysis in a representative ROI showed a noise reduction of 18% in UHR-SPICCS in the BIN-1 () image and 44% in the BIN-2 (65 – 120 keV) image compared to conventional FBP images, while mean CT numbers in the ROIs remained consistent between FBP and UHR-SPICCS images as shown in Table 1.

The MTF measurements for the PCD-CT Macro mode and the UHR mode are shown in Fig 2. The 10% MTF values increased by a factor of two for the UHR mode (0.92 mm^{-1}) compared to the Macro mode (0.45 mm^{-1}). This increase in spatial resolution from the UHR mode images (FBP and UHR-SPICCS) can be attributed to the smaller detector pixel size (0.25 mm at isocenter) incorporated in the PCD-CT's UHR mode. It is to be noted that UHR-SPICCS did not cause alterations in the system spatial resolution. As shown in Figure 2, the MTF values from UHR-SPICCS (solid magenta curve) showed good agreement with those obtained from conventional FBP reconstruction (FDK-UHR) using the same dataset (dashed black curve).

The multi-energy reconstructions for the porcine head scan from conventional FBP and UHR-SPICCS are shown in Fig. 3. The 25 – 65 keV and 65 – 120 keV energy-bin images reconstructed with UHR-SPICCS showed 25% and 46% noise reduction respectively, when

compared to conventional FBP images. Also, the UHR-SPICCS images demonstrated slightly lower noise than the full spectrum prior image as shown in Table 2.

4. CONCLUSIONS

This study demonstrated a multi-energy reconstruction method using the SPICCS framework to achieve low-noise energy-bin image reconstructions for ultra-high resolution PCD-CT imaging. The UHR-SPICCS method did not cause any adverse spatial or spectral alterations compared to conventional FBP images, while achieving 18 – 46% noise reduction in the bin images for the anthropomorphic phantom and swine studies. The high spatial redundancy between energy bin images and the full spectrum acquisition was exploited to achieve ultra-high resolution, low noise multi-energy reconstructions. This approach could be used to generate PCD-CT images for quantitative analysis such as material decomposition and monoenergetic image synthesis that are highly sensitive to image noise.

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Fig 1.

Head phantom reconstructions from the same UHR dataset using FBP and UHR-SPICCS.

The bin images reconstructed using UHR-SPICCS showed lower image noise than conventional FBP reconstructions. CT number and image noise was measured in a representative ROI (shown in the prior image). Please refer Table-1 for CT number and noise measurements in the ROI. W/L = 600/150HU, image size = 512×512 , slice thickness = 0.25 mm.

Fig 2.

MTF measurements from the Macro mode (reproduced from Yu et al⁴) and UHR mode of the PCD-CT system. The spatial resolution curves were similar for FBP (referred as FDK in the plot) and SPICCS. The 10% MTF was 0.92 mm^{-1} for UHR and 0.45 mm^{-1} for Macro mode. This improvement in spatial resolution was expected due to the smaller detector pixel size of the UHR mode.

Fig 3.

FBP and UHR-SPICCS reconstructions from the porcine head scan. The bin images reconstructed using UHR-SPICCS showed lower image noise than conventional FBP reconstructions (see Table 2). CT number and image noise was measured in a representative ROI shown in the prior image. W/L = 600/100HU, image matrix size = 512×512 , slice thickness = 0.25 mm.

Anthropomorphic head phantom: CT number consistency and noise reduction from UHR-SPICCS for the ROI shown in Fig 1.

Table 1

	25 – 120 keV		25 – 65 keV		65 – 120 keV	
	FBP	UHR-SPICCS	FBP	UHR-SPICCS	FBP	UHR-SPICCS
Mean (HU)	147	-	141	139	152	151
Noise (HU)	72	-	87	71	107	60

Porcine head scan: CT number consistency and noise reduction from UHR-SPICCS for the ROI shown in Fig 3.

Table 2

	25 – 120 keV		25 – 65 keV		65 – 120 keV	
	FBP	UHR-SPICCS	FBP	UHR-SPICCS	FBP	UHR-SPICCS
Mean (HU)	41	-	43	42	39	39
Noise (HU)	103	-	131	98	175	93