Published in final edited form as:

Proc SPIE Int Soc Opt Eng. 2019 June ; 11072: . doi:10.1117/12.2534897.

Low-dose photon counting CT reconstruction bias reduction with multi-energy alternating minimization algorithm

Jingwei Lu^a, Shuangyue Zhang^a, David G. Polite^b, Joseph A. O'Sullivan^a

^aWashington University in St. Louis, Department of Electrical and Systems Engineering, 1 Brookings Dr, St. Louis, MO 63130 USA

^bWashington University School of Medicine, Electronic Radiology Laboratory, Mallinckrodt Institute of Radiology, St. Louis, MO 63110 USA

Abstract

Photon counting CT (PCCT) is an x-ray imaging technique that has undergone great development in the past decade. PCCT has the potential to improve dose efficiency and low-dose performance. In this paper, we propose a statistics-based iterative algorithm to perform a direct reconstruction of material-decomposed images. Compared with the conventional sinogram-based decomposition method which has degraded performance in low-dose scenarios, the multi-energy alternating minimization algorithm for photon counting CT (MEAM-PCCT) can generate accurate material-decomposed image with much smaller biases.

Keywords

photon counting; low-dose radiation; iterative algorithm; spectral x-ray CT; alternating minimization

1. INTRODUCTION

Photon counting computed tomography (PCCT) is a promising technology for diagnostic and therapeutic purposes in medical imaging. Compared with conventional CT systems, photon counting systems can provide additional energy-dependent information which enables accurate treatment planning in radiation oncology,¹ material decomposition,² reduced radiation dose,³ reduced beam hardening effects, and higher signal-to-noise ratio.⁴

The most common sinogram-based PCCT material decomposition algorithms use separate steps of material decomposition and image reconstruction.⁵ Separating these steps has the flaw that the geometry and statistical information are not fully used. With decreased radiation dose, the material decomposition has dramatically increased bias.

In this paper, we present a joint statistical reconstruction algorithm called multi-energy alternating minimization (MEAM) algorithm for PCCT systems, which directly reconstructs the material-based images from the measurements. In the patient phantom simulation,

MEAM can effectively maintain small bias even in ultra-low-dose scenarios and reduce the noise with appropriate regularization.

2. METHOD

2.1 Sinogram-based decomposition

Let $d_k(y)$ denote the transmission measurement of energy bin k at the y th source-detector pair. We model the data as having a mutually independent Poisson distribution with mean

$$Q_k(y) = I(y) \int_E D_k(E) S(E) e^{-M(y,E)} dE, \quad (1)$$

where $I(y)$ is the unattenuated photon intensity at the y th source-detector pair, $D_k(E)$ is the bin response function (BRF) of energy bin k , $S(E)$ is the normalized spectrum of the x-ray source, and $M(y, E)$ is the integrated linear attenuation at the y th source-detector pair at energy E .

With the basis vector model (BVM),⁶ we approximate each scanned material as a linear combination of J basis materials with their respective linear attenuation coefficient (LAC), $\mu_j(E)$, or

$$M(y, E) = \sum_{j=1}^J \mu_j(E) l_j(y), \quad (2)$$

where $\mu_j(E)$ is the linear attenuation coefficient of basis material j at energy E and $l_j(y)$ is the line integral length of material j at the y th source-detector pair. Assuming the $d_k(y)$'s are independently Poisson distributed with mean $Q_k(y)$, we estimate the basis-material line integral $l_j(y)$ by minimizing the negative log-likelihood with arbitrary measurement result $d_k(y)$ according to

$$\min_{\{l_j(y), j=1,2,\dots,J\}} L(d_1(y), \dots, d_K(y) | l_j(y)) = \sum_{k=1}^K [Q_k(y) - d_k(y) \log Q_k(y)], \quad (3)$$

where $Q_k(y)$ is defined in (1). By solving problem (3) independently for every y , we decompose the measurements into basis-material sinograms. The conventional filtered backprojection (FBP) algorithm is then used to generate the corresponding basis-material images $\{c_j(x), j=1, 2, \dots, J\}$.

2.2 Multi-energy alternating minimization algorithm

Sinogram-based decomposition methods are easy to implement. However, we found that they may produce errors in low-dose cases (which will be discussed in the next section below). To solve this issue, we developed the MEAM algorithm to reconstruct the basis-material images directly from the transmission data.

In order to directly reconstruct the basis-material images c , we rewrite (2) as

$$M(y, E) = \sum_{j=1}^J \mu_j(E) \sum_x h(y|x) c_j(x), \quad (4)$$

where $h(y|x)$ is a CT system matrix element, and $c_j(x)$ is the basis-material image pixel. It has been proven that the original minimization problem is equivalent to a double-minimization problem

$$\begin{aligned} \min_{p,q} F(p, q) &= \sum_y \sum_k \sum_E p_k(y, E) \log \frac{p_k(y, E)}{q_k(y, E)} - p_k(y, E) + q_k(y, E), \text{ s.t. } \sum_E p_k(y, \\ E) &= d_k(y) q_k(y, E) = \sum_E I(y) D_k(E) S(E) e^{-\sum_{j=1}^J \mu_j(E) \sum_x h(y|x) c_j(x)}. \end{aligned} \quad (5)$$

By solving this problem, we can directly reconstruct the basis-material image c .

3. PHANTOM SIMULATION

3.1 Simulation setup

In this paper, the photon counting detector (PCD) spectral response function (SRF) and bin response function (BRF) were generated using the Photon Counting Toolkit (PcTK),⁷ where the pulse pileup effect is ignored, while the charge sharing effect and the electronic noise are taken into consideration. The radius of electronic charge cloud $\gamma_0 = 24 \mu\text{m}$, the electronic noise parameter $\delta_e = 4 \text{ keV}$, the PCD size in channel-row $d_{pix} = 225 \mu\text{m}$, and the PCD thickness $d_z = 1600 \mu\text{m}$. Measurements with five bins were generated with thresholds $E_1 = 20 \text{ keV}$, $E_2 = 40 \text{ keV}$, $E_3 = 55 \text{ keV}$, $E_4 = 70 \text{ keV}$, and $E_5 = 90 \text{ keV}$. The BRF of these five bins are shown in Figure 1. We tested the accuracy of the sinogram-based decomposition method and the MEAM algorithm with a virtual patient phantom simulation. A 120 kVp source spectrum was used and the patient phantom was a pelvic abdominal slice from the International Commission on Radiological Protection (ICRP) adult female phantom.⁸ The proprietary geometry of the Philips Brilliance Big Bore CT scanner (Philips Medical Systems, Cleveland, OH) with 816 detectors per detector row and 1320 source positions per rotation was used in this simulation. Consistent with previous dual-energy CT analyses from our group, we set $J=2$ and choose polystyrene and 23% calcium chloride aqueous solution as the two basis materials.⁶

To quantitatively compare the accuracy of the sinogram-based decomposition method and the MEAM algorithm, the linear attenuation coefficient images were reconstructed with the BVM,

$$\mu(x, E) = \sum_{j=1}^J \mu_j(E) c_j(x). \quad (6)$$

Then the linear attenuation coefficient images are converted to CT numbers for display.

3.2 Simulation results

Figure 2 shows the ground truth image and reconstructed images of the two methods from 100 mAs simulated data. To quantitatively study the accuracy of the sinogram-based decomposition method with various dose levels, three homogeneous regions of interest (ROIs) are selected, which are highlighted by the red boxes in Figure 2(a). Figure 3 shows the average biases of each selected ROI with dose levels 200 mAs, 100 mAs, 50 mAs, and 30 mAs, respectively. We can see that with the dose level decreasing, the average bias dramatically increases.

The simulation results using the MEAM algorithm are shown in Figure 4. We can see that with decreased dose level, the average bias is strictly restrained to within a very tight window of several HUs. Even in the ultra-low-dose scenario of 30 mAs, where the conventional sinogram-based method has a dramatic positive bias, MEAM can still limit the bias to within a small window around 0.

3.3 Discussion

The major reason for the biased estimation in the sinogram-based decomposition method is that the basis-material line integral $I(y)$ is reconstructed by solving problem (3) independently for every source-detector pair y . However, the assumption that each source-detector pair is independent is not valid, because the source-detector pair ray paths in different scanning angles may intersect, so that the two-step sinogram-based decomposition method is ill-posed. In the MEAM algorithm, we incorporate the forward-projection model, and directly reconstruct the basis-material images $c(x)$ from the statistical model.

4. CONCLUSIONS

The conventional sinogram-based method suffers large bias from the biased basis-material sinogram estimations when the radiation dose is low. The reason is that, in the material decomposition step of the sinogram-based method, we assume the basis-material line integral of every source-detector pair is independent; however, this assumption is not valid because of the PCCT system geometry. We presented the new multi-energy alternating minimization algorithm that reconstructs the basis-material images directly from the transmission data. Our simulation results show that the MEAM algorithm can effectively eliminate the systematic bias in the sinogram-based decomposition method even in ultra-low-dose scenarios.

ACKNOWLEDGMENTS

This study is supported by NIH R01 CA 149305, NIH R01 CA 212638, and NIH 5T32EB01485505.

REFERENCES

- [1]. Zhang S, Han D, Williamson JF, Zhao T, Polite DG, Whiting BR, and O'Sullivan JA, "Experimental implementation of a joint statistical image reconstruction method for proton stopping power mapping from dual-energy CT data," *Medical Physics* 46(1), 273–285 (2018). [PubMed: 30421790]

- [2]. Alvarez RE and Macovski A, "Energy-selective reconstructions in x-ray computerised tomography," *Physics in Medicine & Biology* 21(5), 733 (1976). [PubMed: 967922]
- [3]. Yuan R, Shuman WP, Earls JP, Hague CJ, Mumtaz HA, Scott-Moncrieff A, Ellis JD, Mayo JR, and Leipsic JA, "Reduced iodine load at CT pulmonary angiography with dual-energy monochromatic imaging: comparison with standard CT pulmonary angiography—a prospective randomized trial," *Radiology* 262(1), 290–297 (2012). [PubMed: 22084206]
- [4]. Yu L, Christner JA, Leng S, Wang J, Fletcher JG, and McCollough CH, "Virtual monochromatic imaging in dual-source dual-energy CT: Radiation dose and image quality," *Medical physics* 38(12), 6371–6379 (2011). [PubMed: 22149820]
- [5]. Schlomka J, Roessl E, Dorscheid R, Dill S, Martens G, Istel T, Bäumer C, Herrmann C, Steadman R, Zeitler G, et al., "Experimental feasibility of multi-energy photon-counting K-edge imaging in pre-clinical computed tomography," *Physics in Medicine & Biology* 53(15), 4031 (2008). [PubMed: 18612175]
- [6]. Williamson JF, Li S, Devic S, Whiting BR, and Lerma FA, "On two-parameter models of photon cross sections: Application to dual-energy CT imaging," *Medical physics* 33(11), 4115–4129 (2006). [PubMed: 17153391]
- [7]. Taguchi K, Stierstorfer K, Polster C, Lee O, and Kappler S, "Spatio-energetic cross-talk in photon counting detectors: Numerical detector model (Pc TK) and workflow for CT image quality assessment," *Medical physics* 45(5), 1985–1998 (2018). [PubMed: 29537627]
- [8]. Menzel H, Clement C, and DeLuca P, "ICRP publication 110. Realistic reference phantoms: an ICRP/ICRU joint effort. A report of adult reference computational phantoms," *Annals of the ICRP* 39(2), 1 (2009).

Figure 1:
The bin response functions of the simulated photon counting detector.

Figure 2:

(a) is the ground truth image of CT numbers at 60 keV, (b) is the reconstructed image at 60 keV with the sinogram-based decomposition method, and (c) is the reconstructed image at 60 keV with MEAM. The dose level is 100 mAs. The viewing windows for all images are $[-300\ 300]$ HU.

Figure 3:

The average CT number biases of ROIs with different dose levels: (a) 200 mAs, (b) 100 mAs, (c) 50 mAs, (d) 30 mAs.

Figure 4:

The average CT number biases of ROIs with different dose levels: (a) 200 mAs, (b) 100 mAs, (c) 50 mAs, (d) 30 mAs.