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A simulation study of high-resolution x-ray computed tomography imaging using irregular sampling with a photon-counting detector

Seungwan Lee, Yu-Na Choi, Hee-Joung Kim*

Department of Radiological Science, College of Health Science, Yonsei University, 1 Yonseidae-gil, Wonju, Gangwon-do 220-710, Republic of Korea

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ABSTRACT

The purpose of this study was to improve the spatial resolution for the x-ray computed tomography (CT) imaging with a photon-counting detector using an irregular sampling method. The geometric shift-model of detector was proposed to produce the irregular sampling pattern and increase the number of samplings in the radial direction. The conventional micro-x-ray CT system and the novel system with the geometric shift-model of detector were simulated using analytic and Monte Carlo simulations. The projections were reconstructed using filtered back-projection (FBP), algebraic reconstruction technique (ART), and total variation (TV) minimization algorithms, and the reconstructed images were compared in terms of normalized root-mean-square error (NRMSE), full-width at half-maximum (FWHM), and coefficient-of-variation (COV). The results showed that the image quality improved in the novel system with the geometric shift-model of detector, and the NRMSE, FWHM, and COV were lower for the images reconstructed using the TV minimization technique in the novel system with the geometric shift-model of detector. The irregular sampling method produced by the geometric shift-model of detector can improve the spatial resolution and reduce artifacts and noise for reconstructed images obtained from an x-ray CT system with a photon-counting detector.

1. Introduction

Photon-counting detectors with energy-discrimination capability have been developed for hard X-ray and computed tomography (CT) imaging. Major benefits of photon-counting detectors are to reduce radiation dose and image noise because these detectors have energy thresholds and are based on high-density materials that have high quantum efficiency [1–3]. Photon-counting detectors can be operated above 20–30 keV because the electronic noise, which is caused by the detector leakage current, presents at low energy values [4,5]. They also measure the photon energy deposited by each event and provide the spectral information [6–9]. Although photon-counting detectors with energy-discrimination capability have these advantages, they are hardly used for medical applications.

Loss of total counts, which occurs due to the high flux of photons and limited readout speed of detector electronics, decreases the signal-to-noise ratio (SNR) and uniformity in x-ray tomographic images [10,11]. This effect also results in decreased detection efficiency and inaccurate recording of photons with higher or lower energies than true values. Charge sharing is caused by small pixel size, charge diffusion into adjacent pixels, and non-K-shell x-rays [2,12,13]. Charge from a single-photon absorption in a detector material is shared between neighboring pixels, and this effect results in multiple entries in energy spectrum at lower energies than the true value. Thus, charge sharing increases low energy counts and causes the spectral distortion.

To overcome these limitations, several research groups have studied the analytic modeling of the loss of total counts and correction of charge sharing using algorithms [7,14–16]. Novel chips and electronics, which have improved readout speed and energy resolution, for photon-counting detectors have been also developed [17–19]. A few prototypes of photon-counting detectors with energy-discrimination capability use the pixel binning, which sums up the contents of $n \times n$ pixels, to improve the SNR, reduce the overall size of data, and compensate for the charge sharing [20,21]. Despite these efforts, there still remains a difficulty in using x-ray images generated by the photon-counting detectors for medical diagnosis. The pixel sizes of several photon-counting detectors are larger than that of conventional detectors, which are used for the x-ray imaging in medical fields, due to additional circuits and physical limitations such as loss of total counts and charge sharing [13,20,22,23]. The large pixel size and use of pixel binning lead to reduction in the spatial resolution because the sampling frequency greatly affects the intrinsic resolution of an imaging system even though the quarter shift of detector and the flying focal spot (FFS) can improve the sampling density of system.

Irregular sampling based on a spiral pattern has been studied as a possible solution to improve the spatial resolution and speed

* Corresponding author. Tel.: +82 33 760 2983; fax: +82 33 760 2562.
E-mail addresses: lswan@yonsei.ac.kr (S. Lee), hjk1@yonsei.ac.kr (H.-J. Kim).

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of scanning in the conventional magnetic resonance imaging (MRI) system [24,25]. In this study, we proposed a novel sampling method based on a spiral pattern to improve the spatial resolution and degrade the artifacts caused by under-sampling in x-ray tomographic images with a photon-counting detector which has a large pixel. A micro-CT system based on a photon-counting detector was modeled with analytic and Monte Carlo simulation tools. The raw-data sets, which were acquired with the conventional and novel sampling methods, were reconstructed using filtered back-projection (FBP), algebraic reconstruction technique (ART), and total variation (TV) minimization algorithms. The reconstructed images were compared in terms of normalized root-mean-square error (NRMSE), full-width at half-maximum (FWHM), and coefficient-of-variation (COV).

2. Materials and methods

2.1. Irregular sampling pattern based on the Archimedean spiral

The Archimedean spiral is the locus of points corresponding to the locations over time of a point moving away from a fixed point with a constant speed along a line which rotates with a constant angle [26]. The general Archimedean spiral is described by the following equation in polar coordinates:

$$r = a + b\theta^{1/k}$$

where r is the radial distance, a and b are the constants, θ is the polar angle, and $k=1$. a and b control the starting point of spiral and distance between successive rotations, respectively. We modified the general Archimedean spiral in order to increase the number of samplings in the radial direction using the geometric shift-model of detector as shown below:

$$r_{ij} = a_j + \left(\left(b\sum_{i=1}^{N} i\right)/(Nc)\right)\theta_i{}^{1/k}$$

where a_j is the initial location of a pixel j, b is the size of detector, N is the total number of angular samples (the total number of projections), c is the constant, and θ_i is the ith scan angle. The second term, $((b\Sigma_{i=1}^{N} i)/(Nc))$, controls the distance of detector shift toward one direction at each angle i. The schematic view of the geometric shift-model of detector is described in Fig. 1. This method can produce an irregular sampling pattern based on the Archimedean spiral and improve the sampling density on a detector plane.

Fig. 2 shows the sampling patterns generated by six pixels at each angle with varying distances of detector-shifting on Radon space and the positions of the pixels on a plane after being shifted by the movement of detector. (Radon space is a coordinate determined by a radial distance, r, from a fixed point and an angle, θ, from a fixed direction. The data is acquired in Radon space by line-integral projection) [27]. Compared to the conventional method, the number of samplings in the radial direction increased when the irregular sampling method was applied to the system. The gap between shifted pixels and sampling area increased as a function of the distance of detector-shifting.

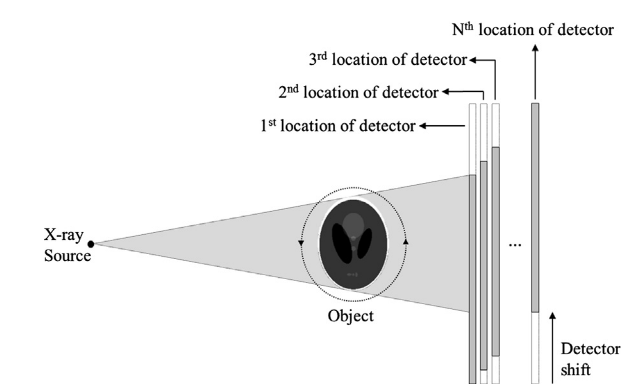

Fig. 1. Schematic view of the geometric shift-model of detector to produce an irregular sampling pattern based on the Archimedean spiral. The detector is moved a constant distance at each angle toward one direction.

2.2. Analytic and Monte Carlo simulations

The micro-x-ray systems with the conventional sampling method and irregular sampling method based on the Archimedean spiral were modeled by using MATLAB version 7.7 (MathWorks Inc., USA). The systems were simulated with a focus-to-center of rotation distance of 900 mm and a focus-to-detector distance of 1000 mm, and the Shepp-Logan phantom, which has an array of 128×128 voxels, was used for the analytic simulation [28]. The phantom was degraded by Gaussian noise with a mean value of the phantom and a standard deviation of 0.01 to obtain noisy projections [29]. The detector had a size of $51.2 \times 0.8\ \mathrm{mm}^2$, an array of 32×1 pixels, and a pixel size of $1.6 \times 0.8\ \mathrm{mm}^2$. To simulate the geometric shift-model of detector, the detector was moved 0.16 mm at each angle toward one direction.

Geant4 Application for Tomographic Emission (GATE) version 6.1 was used to perform the Monte Carlo simulation. GATE is a generic simulation platform based on general-purpose Monte Carlo code Geant4 and advanced open-source software developed by the international OpenGATE collaboration [30]. In this study, the CT scanner module, which simulates interactions between the x-ray photons and the detector materials and displays the data of the attenuated x-ray photons, was used to obtain projections. Electromagnetic processes, such as the photoelectric effect, Compton scattering, Rayleigh scattering, and bremsstrahlung, were taken into account to simulate the electromagnetic interactions of particles with matter. To ensure an efficient simulation, the variation reduction technique, which was developed to decrease simulation time, was used. This technique splits the photons that reach the detector surface into clones, and the simulation time can be decreased by avoiding a new generation and propagation of photons [31].

A mathematical mouse phantom, MOBY phantom, which had a size of $25.984 \times 25.984 \times 81.2\ \mathrm{mm}^3$ and an array of $128 \times 128 \times 400$ voxels, was used for the Monte Carlo simulation. MOBY phantom is a realistic and flexible 4-dimensional digital mouse phantom for small animal imaging in simulation studies [32]. This phantom includes various organs and models the complex anatomy of different organs. The photon-counting detector was modeled on an eValuator™-2500 (eV-Products, USA) linear array sensor based on cadmium zinc telluride (CZT). It has a size of $51.2 \times 0.8\ \mathrm{mm}^2$, an array of 64×1 pixels, a pixel pitch of $0.8 \times 0.8\ \mathrm{mm}^2$, and a thickness of 2 mm. This detector can be operated in the energy range of 20–160 keV, and the count rate of this detector is above 5×10^5 counts per second per pixel (cps/pixel). In this study, the charge sharing was not taken into account in the simulation. As is the case with the analytic simulation, the detector was moved 0.16 mm at each angle to simulate the geometric shift-model of detector, and the data was obtained with an array of 32×1 pixels which summed up the contents of 2×1 pixels. In practice, the input flux of photon-counting detector must be below 5×10^5 counts per second per square millimeter (cps/mm^2) in order to avoid the spectral distortion due to the pulse pile-up effect [10]. The incident x-ray spectrum was simulated at 60 kVp and 50 μA with 0.8 mm beryllium (Be), 10 mm aluminum (Al), and 0.11 mm copper (Cu) filtrations using the SpekCalc program (REAL software Inc., USA) [33]. The simulated spectrum