Transactions of the Korean Nuclear Society Autumn Meeting
Gyeongju, Korea, October 24-25, 2013

Analytical simulation platform describing projections in computed tomography systems

Hanbean Youn and Ho Kyung Kim*
School of Mechanical Engineering, Pusan National University, Busan 609-735, South Korea
Center for Advanced Medical Engineering Research, Pusan National University, Busan 609-735, South Korea

**Corresponding author: hokyung@pusan.ac.kr*

1. Introduction

Recent developments in computed tomography (CT) systems are focused on the patient dose. To reduce the patient dose, several approaches such as spectral imaging using photon counting detectors and statistical image reconstruction, are being considered. Although image-reconstruction algorithms may significantly enhance image quality in reconstructed images with low dose, true signal-to-noise properties are mainly determined by image quality in projections.

We are developing an analytical simulation platform describing projections to investigate how quantum-interaction physics in each component configuring CT systems affect image quality in projections. This simulator will be very useful for an improved design or optimization of CT systems in economy as well as the development of novel image-reconstruction algorithms. In this study, we present the progress of development of the simulation platform with an emphasis on the theoretical framework describing the generation of projection data.

2. Methods and Results

2.1 Simulation platform

We first prepared simulation platform containing conventional ray tracing algorithm proposed by Siddon[1]. Siddon's method is a typical line-weighted approach, which regards voxels as sets of sub-planes and calculates interacting points through the trajectory including source and pixels. Thus we defined positions of source and pixels with respect to the helical motion using the universal coordinates to describe curved detector, rotational motion of gantry and translational motion of patient table.

2.2 Imaging chain

In order to consider image noise in projections or the random nature in pixel-by-pixel signals, we developed the cascaded signal-transfer model as described in Fig. 1. Assuming that all interaction processes in signal formation in CT systems follow the Poisson statistics, the projection model includes several random signal propagation stages: attenuation through objects $N_a(E)$, x-ray detection by scintillators $\alpha(E)$, secondary quantum gain in scintillators (i.e., x-ray-to-light conversion) $\beta(E)$, light detection in photodiodes (i.e., light-to-electronic charge conversion) η, and the addition of electronic noise quanta n_{elec}. We note that the randomness in each interaction stage is considered in pixel-by-pixel ways in projections.

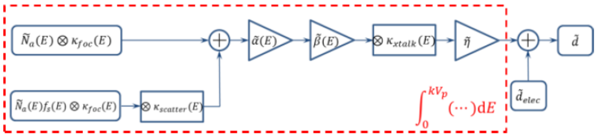

Fig. 1. Cascaded signal-transfer model describing pixel signals in projections of CT systems. The overhead tilde designates a random variable. The symbol "⊗" is the convolution operator. The large dotted box denotes iterative signal propagation over given x-ray spectrum.

To account for image-signal blur in projections, we included deterministic blur stages due to finite-sized focal spot, x-ray scatter in objects, and light signal spreading in pixels, also called the pixel crosstalk, in the projection model. The convolution kernel due to the focal spot blurring was obtained by the volume-weighted integration in ray tracing or by using the transfer function assuming the Gaussian spatial distribution of beam intensities in the focal spot. The difference between two analytical approaches was, however, negligible. On the other hand, the convolution kernels describing x-ray scatter and light signal crosstalk in detectors were obtained from the Monte Carlo simulations. MCNP (RSICC, Oak Ridge, TN, USA) and DETECT2000 (Laval University, Quebec, Canada) were used for x-ray and light photon transports, respectively.

2.3 Description of the Simulator

Coordinates used in the CT projection simulator were described as cylindrical coordinates. The simulator included cone-beam projection, helical motion of multi-channel detectors, and misalignments among the source, the object and the detector. All the geometrical motions were achieved by matrix operations, such as rotation and translation.

We took computer-generating x-ray spectra for the poly-energetic source sampling [2], [3]. For the Radon transform of voxelized phantom, we first adapted Siddon's method and updated the ray-tracing method with Gao's sorting method [4] that was a modified version of the conventional Siddon's method [1]. The x-ray beam filtration and attenuation were performed based on the attenuation coefficients from the National Institute of Standards and Technology. In case of the

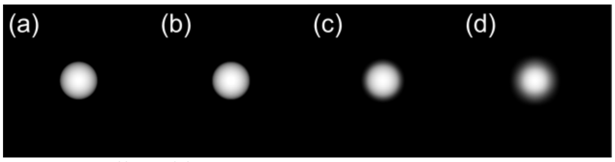

Fig. 2. The effect of focal spot blur at the imager. (a)-(d) are the cases of 0.1, 0.2, 0.5 and 1 mm focal spots.

bowtie filtration, we adapted Boone's semi empirical model [5].

2. Results

Figure 2 demonstrates the effect of focal spot blur in cases of 0.1, 0.2, 0.3 and 0.5 mm focal spots. The effect of focal spot can be predicted using the Gaussian shaped transfer functions. Figure 3 shows the effect of poly-energetic sampling for a voxel chest phantom, which was extracted from a patient data scanned by helical CT, with a format of 512 × 512 × 675 voxels in 0.617 × 0.607 × 3.2-mm voxel size under irradiation of 120-kV tungsten spectrum. In this simulation, we considered curved multi-channel detector array with 128 channels and 900 pixels per row in which each pixel size was assumed to 1.2 mm. The average ionization energy of scintillator and quantum efficiency of photodiode were assumed to 17 eV and 0.55, respectively. And we also considered the effect of the patient dose as shown in Fig. 4. All the results are not only compared in projection domain, but also compared in reconstruction domain. More detailed results will be demonstrated.

2. Conclusion

We have prepared the analytical simulation platform describing projections in computed tomography systems. The remained further study before the meeting includes the following:

Each stage in the cascaded signal-transfer model for obtaining projections will be validated by the Monte Carlo simulations.

- We will build up energy-dependent scatter and pixel-crosstalk kernels, and show their effects on image quality in projections and reconstructed images.
- We will investigate the effects of projections obtained from various imaging conditions and system (or detector) operation parameters on reconstructed images.
- It is challenging to include the interaction physics due to photon-counting detectors into the simulation platform.

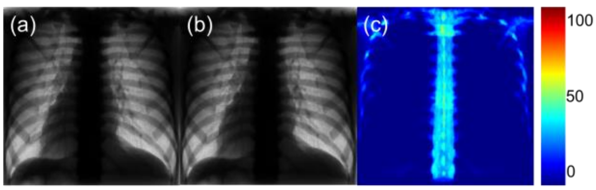

Fig. 3. Comparisons to the poly-energetic and mono-energetic considerations. (a) shows the case of 120 kVp spectrum incidence and (b) shows the 64 keV mono-energy, which is average energy of 120 kVp spectra, incidence. And(c) shows relative errors compared with (a) and (b).

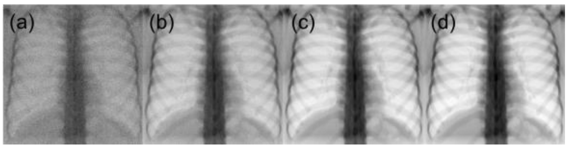

Fig. 4. Noise considered numerical projections along the various incident doses. (a), (b), (c) and (d) are cases of 0.01 mR, 0.1 mR, 1 mR and 10 mR, respectively.

Detailed descriptions of the simulator will be presented with discussions on its performance and limitation as well as Monte Carlo validations. Computational cost will also be addressed in detail.

The proposed method in this study is simple and can be used conveniently in lab environment.

ACKNOWLEDGEMENTS

This work was supported by Samsung electronics and Radiation Technology R&D program through the National Research Foundation of Korea funded by the Ministry of Science, ICT & Future Planning.

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