

A computer simulation method for low-dose CT images by use of real high-dose images: a phantom study

Tomomi Takenaga¹ · Shigehiko Katsuragawa^{2,3} · Makoto Goto⁴ · Masahiro Hatemura⁴ · Yoshikazu Uchiyama² · Junji Shiraiishi²

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Abstract Practical simulations of low-dose CT images have a possibility of being helpful means for optimization of the CT exposure dose. Because current methods reported by several researchers are limited to specific vendor platforms and generally rely on raw sinogram data that are difficult to access, we have developed a new computerized scheme for producing simulated low-dose CT images from real high-dose images without use of raw sinogram data or of a particular phantom. Our computerized scheme for low-dose CT simulation was based on the addition of a simulated noise image to a real high-dose CT image reconstructed by the filtered back-projection algorithm. First, a sinogram was generated from the forward projection of a high-dose CT image. Then, an additional noise sinogram resulting from use of a reduced exposure dose was estimated from a predetermined noise model. Finally, a noise CT image was reconstructed with a predetermined filter and was added to the real high-dose CT image to create a simulated low-dose CT image. The noise power spectrum and modulation transfer function of the simulated low-dose images were very close to those of the real low-dose images. In order to confirm the feasibility of our method, we applied this method to clinical cases which were

examined with the high dose initially and then followed with a low-dose CT. In conclusion, our proposed method could simulate the low-dose CT images from their real high-dose images with sufficient accuracy and could be used for determining the optimal dose setting for various clinical CT examinations.

Keywords Low-dose CT · Simulation · Filtered back-projection · NPS · MTF

1 Introduction

The average typical effective dose in a CT examination is much larger than that in a general radiographic examination, whereas CT examinations provide much useful information for accurate diagnoses. For example, the dose for a chest CT examination is 250–350 times larger than that for chest radiography [1, 2]. Therefore, it is necessary to optimize the exposure dose in CT examinations. Because multiple exposures to the same patient for determining the optimal dose in a CT examination are ethically unacceptable, a number of studies in which low-dose CT simulation was used have been proposed [3–8]. If low-dose CT images are simulated with a high degree of accuracy, we will be able to estimate a personalized exposure dose that is suitable for each individual patient. In this study, we note that “low dose” means a relatively low exposure dose compared to that of the standard diagnostic CT examination, whereas low-dose CT sometimes meant a large reduction of the exposure dose for screening CT. In a low-dose CT examination with a reconstruction algorithm of filtered back-projection (FBP), the increased noise caused by the reduction of the exposure dose is added to the raw sinogram data, and then the noise on a CT image is propagated

✉ Tomomi Takenaga
138w5004@st.kumamoto-u.ac.jp

¹ Graduate School of Health Sciences, Kumamoto University, 4-24-1 Kuhonji, Kumamoto 862-0976, Japan

² Faculty of Life Sciences, Kumamoto University, 4-24-1 Kuhonji, Kumamoto 862-0976, Japan

³ Faculty of Fukuoka Medical Technology, Teikyo University, 6-22 Misakimachi, Omuta, Fukuoka 836-0037, Japan

⁴ Department of Radiology, Kumamoto University Hospital, 1-1 Honjo, Kumamoto 862-0976, Japan

by the FBP reconstruction. Therefore, low-dose CT simulations in previous studies can be classified into two approaches: one based on raw sinogram data [3–5] and the other based on reconstructed CT images [6–8].

The first approach, which was based on raw sinogram data, can simulate realistic low-dose images, because real sinograms acquired by a CT scanner and a reconstruction filter of the CT system can be employed in a simulation study [3–5]. Although this approach is ideal for a low-dose simulation study, it is almost impossible for many users to access raw sinogram data without permission of the CT manufacturers, and thus this approach is limited to specific vendor platforms.

The second approach, which is based on reconstructed CT images, can be classified further into two methods. In one method, the increased noise is added to the reconstructed images directly. In the other method, the increased noise is added to sinograms which were generated from the high-dose CT images, and then the low-dose CT images are simulated by use of an appropriate reconstruction filter. It should be noted that, in this study, the term “high dose” means the ordinary exposure dose for standard CT examinations.

Britten et al. proposed a method that added spatially correlated noise to the high-dose CT image [6]. They assumed the same noise level over the entire CT image; however, the noise is different in different locations depending on the tissues. Li et al. applied a noise addition software tool (GE Noise Addition Tool, GE Healthcare, Waukesha, WI, USA) to simulate low-dose CT images [7]. This software tool generated sinograms from high-dose CT images and estimated additional noise in the sinograms according to the reduced exposure dose. However, because this software is applicable only to CT images obtained with GE scanners, their method is also limited to a specific vendor platform. Kim et al. developed a method for simulation of low-dose CT images without the requirement for raw sinogram data [8]. In their method, the additional noise in the simulated sinograms was estimated by a noise model which was determined from the measurement of the noise-equivalent quanta (NEQ) for reconstructed CT images. Although their method could provide low-dose CT images with a sufficient level of accuracy, a phantom with a particular shape, such as tapered acrylic phantom, was required for NEQ measurement.

Generally speaking, it is necessary for clinical applications to develop a simpler simulation method for further expanding the versatility without use of special equipment. In this study, we propose a novel low-dose CT simulation method with use of only high-dose CT images. Our method is simple and practical, because it does not need the use of a particular phantom or raw sinogram data, and simulated CT images can be provided by use of various existing CT

images which were archived in most of medical institutions and/or hospitals.

2 Methods

2.1 Overall scheme of low-dose CT simulation

The overall scheme of low-dose simulation is shown in Fig. 1. In our simulation method, first a sinogram was generated from the forward projection of a high-dose CT image. Then, an additional noise sinogram resulting from a reduced exposure dose was estimated by a predetermined noise model which will be described in Sect. 2.4. Finally, a noise CT image was reconstructed with a predetermined filter which will be described in Sect. 2.5 and was added to the high-dose CT image to create a simulated low-dose CT image.

As described later, the noise model and the reconstruction filter were determined by use of real high- and low-dose CT images of a given phantom, which is referred to here as a reference phantom. The noise model estimated the magnitude of the noise added to the sinogram, and the reconstruction filter played a role in modifying the frequency distribution of the additional noise on the simulated low-dose CT images.

2.2 Experimental materials

An American College of Radiology (ACR) CT accreditation phantom (model 464, Gammex-RMI, Middleton, WI, USA), a Catphan phantom (CTP486, The Phantom Laboratory, Salem, NY, USA), and a TOS phantom (Toshiba Medical Systems Corporation, Otawara, Japan) were scanned by a 64-slice multidetector-array CT system (Brilliance; Philips Healthcare, Cleveland, OH, USA) with use of the acquisition parameters shown in Table 1. Axial images were reconstructed by the FBP algorithm with a standard reconstruction filter (type C). The tube voltage

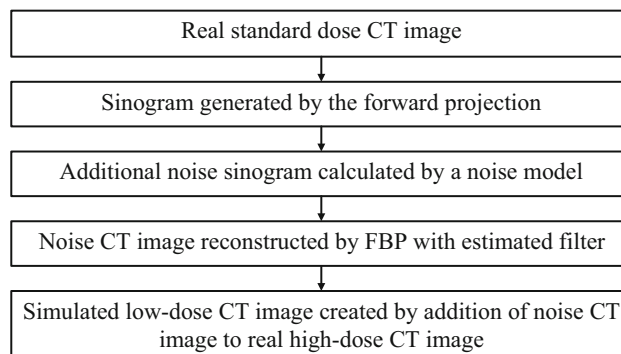


Fig. 1 Summary of the simulation scheme

Table 1 Acquisition parameters used for low-dose CT image simulation

Parameter	Value
Scan type	Axial
Tube voltage (kV)	120
Tube current (mA)	100, 200, 300
Slice thickness (mm)	0.625, 1.250, 2.500, 5.000
Pixel size (mm ²)	0.684 × 0.684
Time per rotation (s)	1.0

and the display field of view were consistently 120 kV and 350 mm, respectively, whereas the tube currents and slice thicknesses were changed for investigation of their effects on the low-dose CT simulation. The three phantoms as shown in Fig. 2 were used as reference phantoms or test phantoms.

2.3 Noise model

Projection values in a sinogram approximately follow a Gaussian distribution with mean p and variance σ^2 [9]:

$$\sigma^2 \propto \frac{1}{d} \cdot e^{-p}, \quad (1)$$

where d is tube current–time product (exposure dose) and p is the attenuation of the object along the X-ray path at a detector. The variance of the projection value σ_{low}^2 for low-dose exposure d_{low} can be expressed as follows:

$$\sigma_{\text{low}}^2 = \sigma_{\text{high}}^2 + \sigma_{\text{add}}^2, \quad (2)$$

where σ_{high}^2 is the variance of the projection value for high dose d_{high} ($>d_{\text{low}}$) and σ_{add}^2 is the variance of additional noise in the sinogram. From Eqs. (1) and (2), we can determine the noise model for the variance of the additional noise in the sinogram corresponding to the simulated low-dose CT image as follows:

$$\begin{aligned} \sigma_{\text{add}}^2 &= \frac{c}{d_{\text{low}}} \cdot e^{-p} - \frac{c}{d_{\text{high}}} \cdot e^{-p} \\ &= \frac{c}{d_{\text{high}} \cdot d_{\text{low}}} \cdot (d_{\text{high}} - d_{\text{low}}) \cdot e^{-p}. \end{aligned} \quad (3)$$

The parameter c is a constant which is the conversion factor for the dose-dependent noise, and p is the attenuation along the X-ray path which is obtained approximately from the sinogram. Therefore, the additional noise sinogram is created by generation of random variables of a Gaussian distribution with zero mean and variance σ_{add}^2 . Note that it is necessary to determine a parameter c in advance for estimating the magnitude of the additional noise in the sinogram and thus for simulating low-dose CT images.

2.4 Determination of parameter in noise model

To determine the parameter c of the noise model shown by Eq. (3), we used CT images obtained from a reference phantom with high dose d_{high} and low dose d_{low} . First, projection data of the sinogram were generated by the forward projection of the high-dose CT image. We varied the parameter c in Eq. (3) and provided various additional noise sinograms for subsequent processing. Then, many additional noise images were reconstructed by FBP with a Ram–Lak filter, which is a basic reconstruction filter derived from the FBP theory in a straightforward manner.

The standard deviation (SD) of the additional noise could be measured from real high- and low-dose CT images as follows:

$$\text{SD}_{\text{add}} = \sqrt{\text{SD}_{\text{low}}^2 - \text{SD}_{\text{high}}^2}, \quad (4)$$

where SD_{low} and SD_{high} are SDs within the region of interest (ROI) in the same location of the CT images with d_{low} and d_{high} , respectively. Finally, the parameter c was determined experimentally in such a way that the SD of the simulated additional noise image reconstructed from the

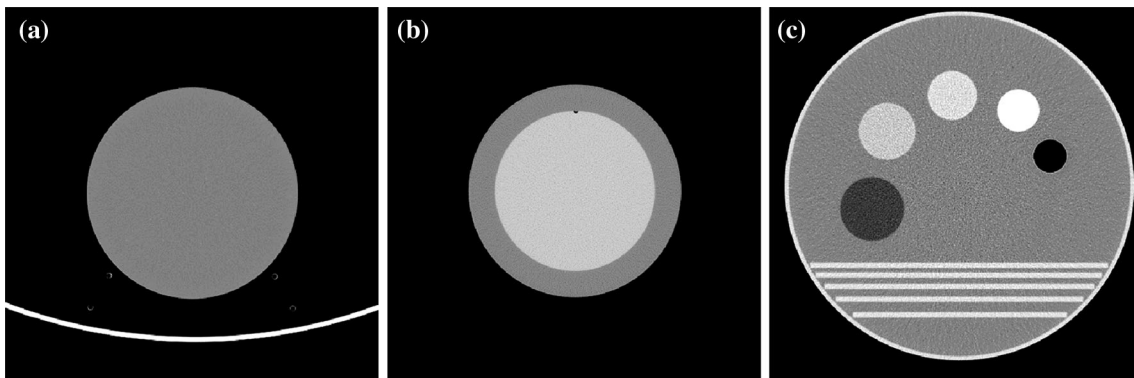


Fig. 2 CT images of the phantoms used in this study: **a** the third module of the ACR phantom, **b** CTP 486 image uniformity module of the Catphan phantom, and **c** TOS phantom. The background material is water equivalent

sinogram generated by Eq. (3) was equal to the SD of the real additional noise measured from Eq. (4). It should be noted that the parameter c was determined from a reference phantom, but the variance of the additional noise on the sinogram in Eq. (3) varies depending on the attenuation p along the X-ray path obtained from any CT images such as clinical images.

2.5 Determination of reconstruction filter

Not only the magnitude but also the frequency distribution of the noise on the simulated low-dose images should match those on the real low-dose images. The frequency distribution of the noise can be modified by use of an appropriate reconstruction filter in the FBP algorithm. Because the noise power spectrum (NPS) of the additional noise image could be measured by the difference between the NPSs of the real CT image with d_{low} and d_{high} , the reconstruction filter for the additional noise image can be formulated as follows:

$$E(f) = \sqrt{\frac{\text{NPS}_{\text{low}}(f) - \text{NPS}_{\text{high}}(f)}{\text{NPS}_{\text{RL}}(f)}} \cdot f, \quad (5)$$

where f is the spatial frequency; $\text{NPS}_{\text{low}}(f)$ and $\text{NPS}_{\text{high}}(f)$ are the NPSs of real CT images with d_{low} and d_{high} , respectively, and $\text{NPS}_{\text{RL}}(f)$ is the NPS of the simulated additional noise image reconstructed with a Ram-Lak filter. Note that the NPS was determined from a two-dimensional Fourier transform (2D-FFT) in an ROI of 128×128 matrix size on CT images of the reference phantom [10].

2.6 Investigation of effect of various factors

For the investigation of the effects of acquisition parameters on the simulated low-dose CT images, we changed high dose d_{high} (200 or 300 mAs) and low dose d_{low} (100 or 200 mAs). The ACR phantom was employed as a reference phantom for determination of a noise model and a reconstruction filter. The noise model and the reconstruction filter were applied for high-dose CT images of the Catphan and TOS phantoms to provide simulated low-dose images. The modulation transfer functions (MTFs) and NPSs of the simulated low-dose CT images were compared with those of the real CT images for evaluation of the effects of the acquisition parameters. MTFs were measured by use of the circular edge method, which utilizes a circular structure in CT images [11]. We used the disk image of a bone-mimicking material with a diameter of 25 mm in the ACR phantom, the polyoxymethylene disk image with a diameter of 40 mm in the TOS phantom, and the cylinder of Catphan with a diameter of 200 mm as circular structures.

The NPSs were measured by use of 2D-FFT in an ROI with 128×128 matrix size, which was located at the image center. The slice thickness was kept at 0.625 mm during investigation of the effects of acquisition parameters.

In addition, we investigated the effects of a reference phantom, exposure dose, and slice thickness on the noise model and the reconstruction filter. In this investigation, we selected the phantom used as a reference phantom from the ACR, Catphan, and TOS phantoms in rotation.

2.7 Preliminary application for clinical images

In order to confirm the feasibility of our method, we applied this method to clinical cases which were examined with the high-dose CT initially and then followed with low-dose CT. Chest CT examinations of 5 patients who had trephine lung biopsies were selected retrospectively from our clinical database obtained by a 64-slice multidetector-array CT (Brilliance; Philips Healthcare, Cleveland, OH, USA). In our hospital, CT examinations with high dose were performed before trephine lung biopsy. Subsequently, CT examinations with a low exposure dose were performed as follow-up examinations after trephine lung biopsy in the effort to check the risk of complications. Therefore, in these particular cases, CT images were obtained with high doses of 105–339 mAs initially, then followed by the use of a low dose of 50 mAs, at a constant tube voltage of 120 kV and a slice thickness of 5 mm. A simulated low-dose (50 mAs) CT image was reconstructed from the real high-dose image acquired before biopsy. At that time, the noise model and the reconstruction filter were obtained from ACR phantom images with a slice thickness of 5.0 mm. Simulated CT images were compared to CT images obtained with a real low exposure dose (50 mAs) in terms of SD within ROIs which were located in the liver parenchyma. All procedures used in this research were approved by the Ethics Committee of Kumamoto University.

3 Results

The parameter c of the conversion factor in the noise model expressed by Eq. (3) was estimated to be 0.00032 mAs from the ACR phantom CT images with a high dose of 300 mAs and a low dose of 100 mAs. In addition, the reconstruction filter was determined from the NPSs of those real CT images and a simulated noise image reconstructed with a Ram-Lak filter, as shown in Fig. 3.

The noise model and the reconstruction filter determined from the ACR phantom were applied to the Catphan and TOS phantoms. The CT images with a low dose of 100 mAs were simulated from the real CT images with a

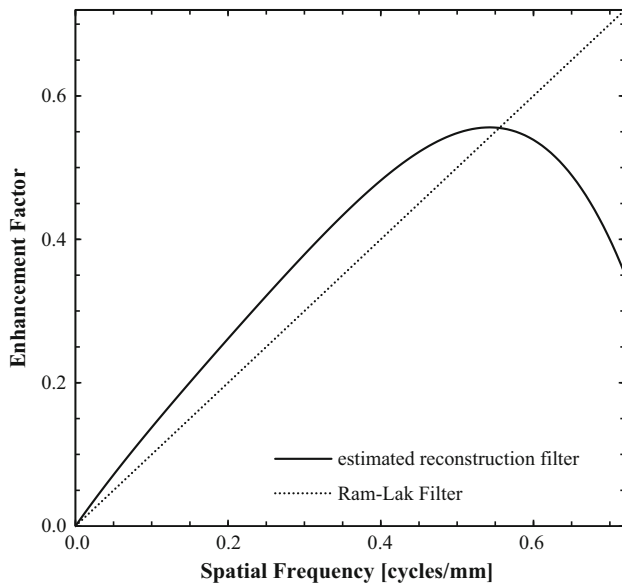


Fig. 3 The estimated reconstruction filter used for the low-dose simulation in this study

high dose of 300 mAs. The noise textures of the simulated low-dose CT images appeared to be similar to those of the real low-dose CT images, as demonstrated in Fig. 4. The noise in terms of the SD of Hounsfield values in an ROI located at the image center with 128×128 matrix size was determined as shown in Table 2. The average SDs of the simulated low-dose images were very close to those of the real low-dose images.

For comparison of the resolution property between real and simulated low-dose images, we measured the MTFs as shown in Fig. 5. The MTFs of the Catphan and TOS phantom images are shown in Fig. 5a, b, respectively. The low-dose CT simulation was carried out for 100 mAs and 200 mAs from 300 mAs, and for 100 mAs from 200 mAs. Although there are slight variations among the MTFs of the TOS phantom, the MTFs of simulated images mostly agreed with those of the real high-dose images. Our results indicate that the resolution property of CT images was not affected by the low-dose simulation for both Catphan and TOS phantoms.

The NPSs obtained from the CT images for the Catphan and TOS phantoms are shown in Fig. 6a, b, respectively. The low-dose CT simulation was carried out under the same dose conditions as in the MTF measurement. Both NPSs of low-dose (100 mAs) images simulated from 200 and 300 mAs were close to the NPS of the real low-dose (100 mAs) image. Our results indicate that the simulated images had a noise property equivalent to that of the real low-dose images for both Catphan and TOS phantoms.

The effect of a reference phantom on the relationship between the parameter c and the slice thickness is shown in Fig. 7a. The high dose and the low dose for the simulation

were 300 and 100 mAs, respectively, in this investigation. The effect of the exposure dose for the simulation is shown in Fig. 7b. Our results indicated that the reference phantom and the exposure dose for the simulation did not affect the parameter c of the noise model very much, whereas the parameter c varied nonlinearly depending on the slice thickness.

The effects of various factors on the reconstruction filter are shown in Fig. 8. The effect of the slice thickness for low-dose simulation of 100 mAs from 300 mAs with the reference ACR phantom is shown in Fig. 8a. The effect of exposure doses with a slice thickness of 0.625 mm and the reference ACR phantom is shown in Fig. 8b. In addition, the effect of reference phantoms for low-dose simulation of 100 mAs from 300 mAs with the slice thickness of 0.625 mm is shown in Fig. 8c. Our results indicated that the reconstruction filter was not affected by the slice thickness, the exposure dose for the simulation, or the reference phantoms.

Figure 9 demonstrates one example of the application of our method to clinical CT images before and after trephine lung biopsy. The simulated low-dose (50 mAs) CT image was reconstructed by use of the real high-dose (322 mAs) image acquired before biopsy. The SDs within ROIs located in the liver parenchyma were 9.6 ± 1.1 , 22.4 ± 2.7 , and 23.2 ± 2.7 in real high-dose, real low-dose, and simulated low-dose CT images, respectively. As shown in Fig. 9, our method could simulate images similar to real low-dose images in a visual sense.

4 Discussion

A computerized scheme for simulation of low-dose CT images from real high-dose images was developed without use of raw sinogram data. The noise textures of the simulated images were visually similar to those of the real images, as shown in Fig. 4. The SDs (Table 2) and NPSs (Fig. 6) of the simulated low-dose images were very close to those of the real low-dose images. In addition, the MTFs of the simulated images were almost the same as those of the real images, as shown in Fig. 5, although there were slight variations among the MTFs of TOS phantom images because of an increase of noises due to a large diameter of the TOS phantom. Therefore, these results indicate that, with our proposed method, we could simulate a low-dose CT image from its real high-dose image with sufficient accuracy.

The advantage of our method is that any CT image, including a clinical image, could be simulated from its real high-dose image by use of the noise model and the reconstruction filter determined from a reference phantom such as the ACR phantom, because the noise model and the

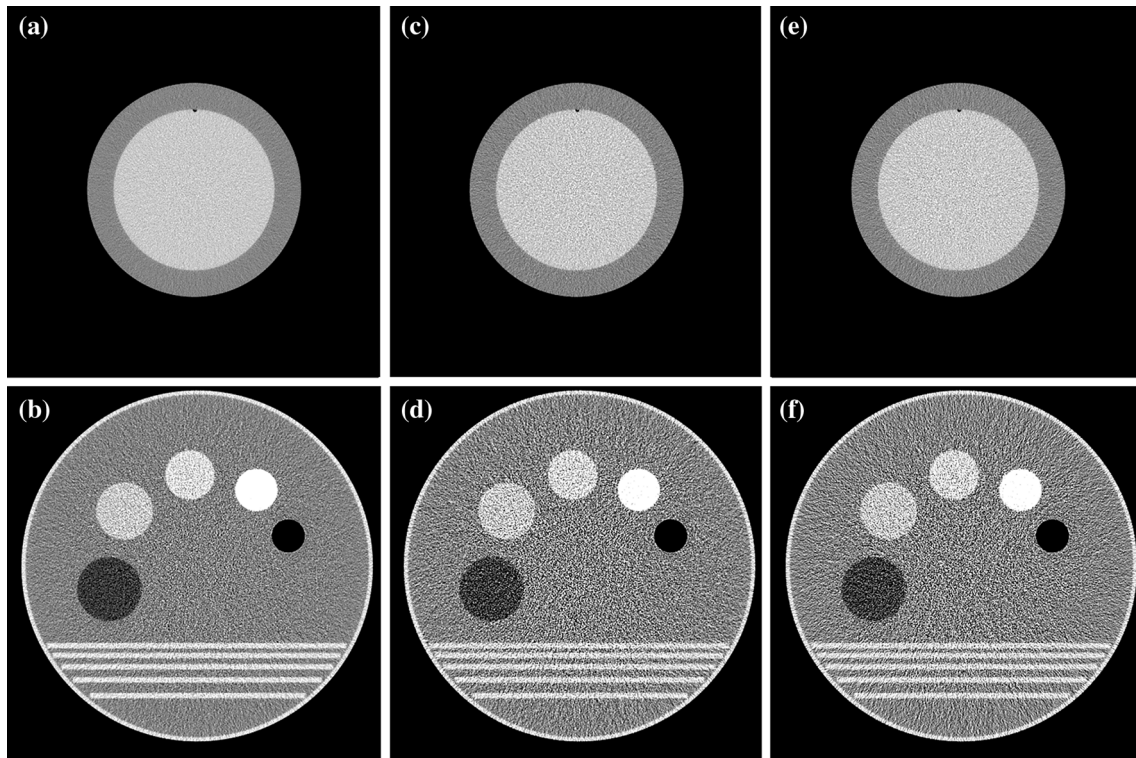


Fig. 4 Demonstration of low-dose simulation for Catphan and TOS phantoms: **a, b** real high-dose CT images (300 mAs); **c, d** real low-dose CT images (100 mAs); and **e, f** simulated low-dose images (100 mAs)

Table 2 Comparison of noise in terms of standard deviation for Catphan and TOS phantoms

Phantom	High-dose image (HU)	Low-dose image (HU)	Simulated low-dose image (HU)
Catphan	15.1 ± 0.2	26.7 ± 0.3	27.5 ± 0.2
TOS	50.7 ± 0.5	89.5 ± 0.7	89.8 ± 0.6

High dose and low dose correspond to 300 and 100 mAs, respectively. The noise model and the reconstruction filter were determined from an ACR phantom

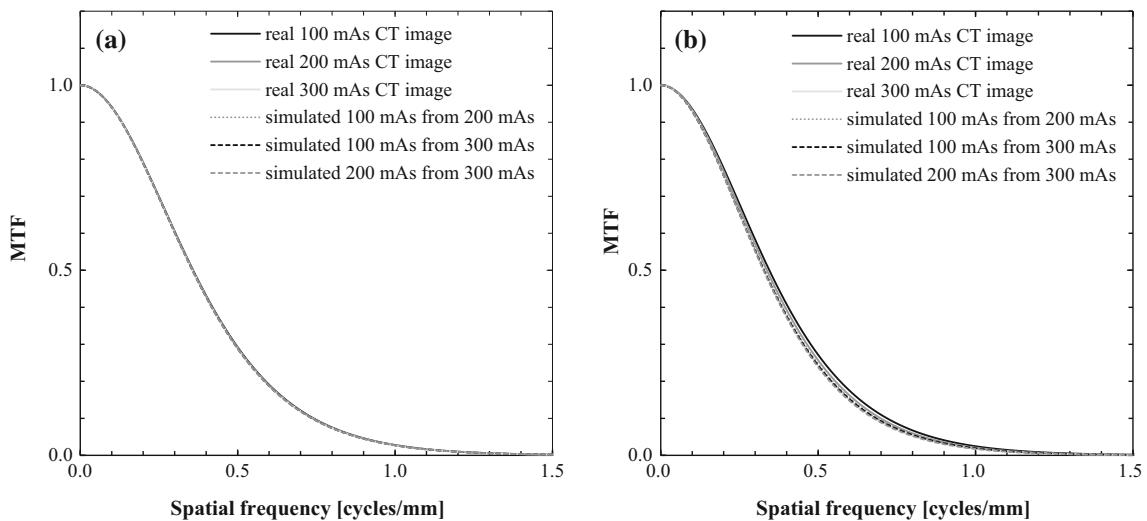


Fig. 5 Comparison of MTFs measured from real CT images and simulated images. **a** Catphan phantom and **b** TOS phantom

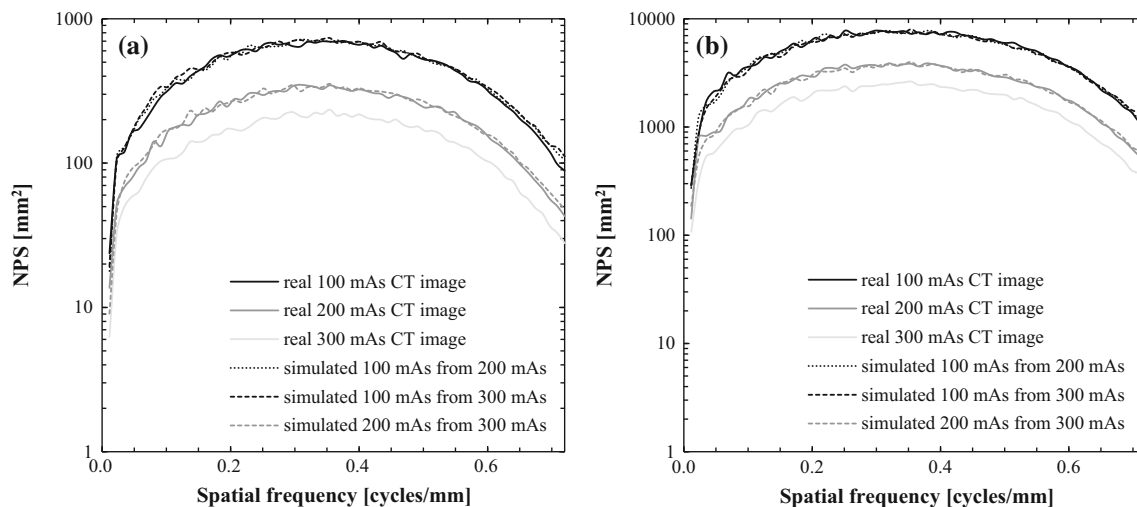


Fig. 6 Comparison of NPSs measured from real CT images and simulated images. **a** Catphan phantom and **b** TOS phantom

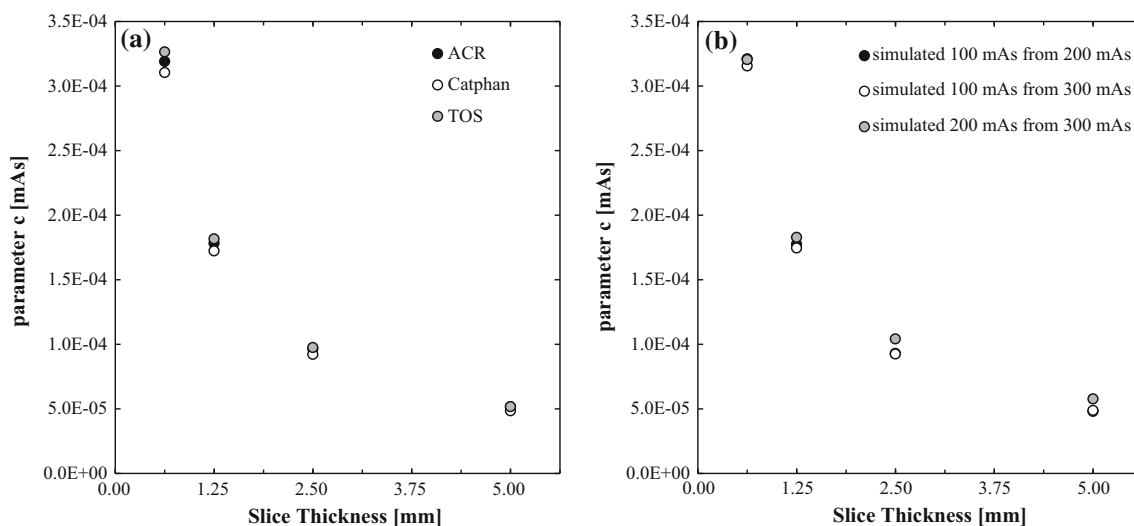


Fig. 7 Relationship between slice thickness and parameter c of the noise model. **a** Effect of reference phantoms. **b** Effect of exposure doses when an ACR phantom was used as a reference phantom

reconstruction filter were not affected by the reference phantom or the exposure dose, as shown in Figs. 7 and 8. However, because the parameter c decreased nonlinearly as the slice thickness was increased, the noise model corresponding to the slice thickness was required for the low-dose simulation. In addition, the reconstruction filter used for the simulation should be varied according to the filter which is employed for the reconstruction of real high-dose CT images. It is not difficult to prepare the noise model and the reconstruction filter according to the slice thickness and the original reconstruction filter, because one needs only any high- and low-dose CT images obtained from a reference phantom.

We did not take account of the electrical noise and the bow-tie filter in making the noise model expressed by Eq. (3), because the noise model was determined from real high- and low-dose CT images which contained the effects of these two factors. Thus, it would appear that the noise model in this study represents the overall additional noise including the electrical noise and the effect of the bow-tie filter.

Our simulated results were compared with previous studies that were based on raw sinogram data. Žabić et al. reported that the SDs of simulated low-dose CT images for a 35-cm water phantom were in error by less than 5% [5]. They simulated CT images with low doses (10–450 mAs)

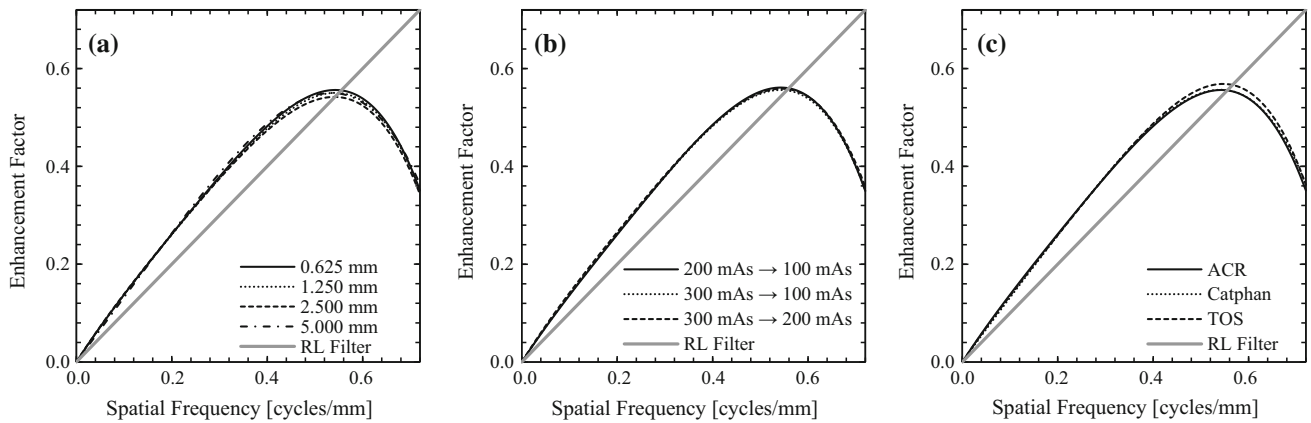


Fig. 8 Effects of various factors on the reconstruction filter. **a** Effect of the slice thickness for low-dose simulation of 100 mAs from 300 mAs with the reference ACR phantom. **b** Effect of exposure

doses with a slice thickness of 0.625 mm and the reference ACR phantom. **c** Effect of reference phantoms for low-dose simulation of 100 mAs from 300 mAs with the slice thickness of 0.625 mm

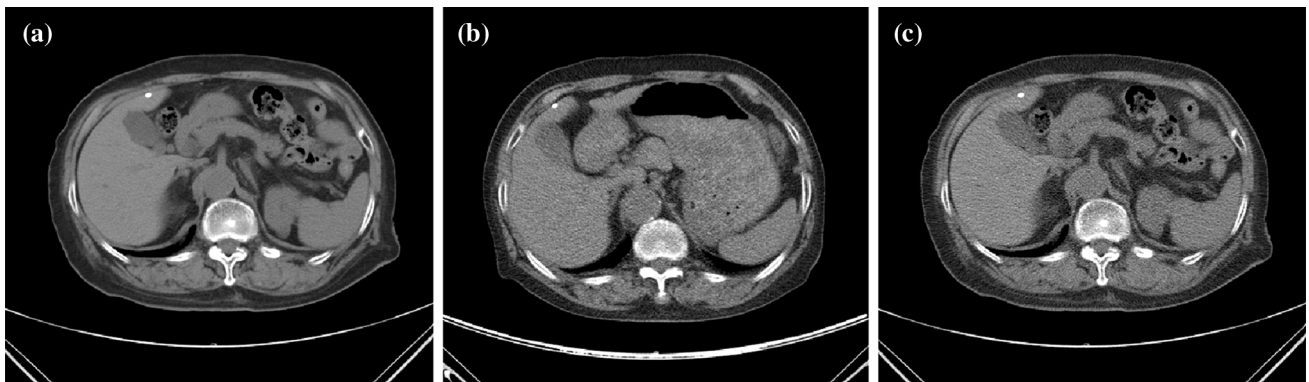


Fig. 9 Demonstration of low-dose simulation for a clinical CT image: **a** real high-dose CT image (322 mAs), **b** real low-dose CT image (50 mAs), and **c** simulated low-dose image (50 mAs)

from a CT image with a high dose (500 mAs). Frush et al. reported that the SDs of simulated low-dose CT images for a 25-cm water phantom were in error by less than 2 % [3]. They simulated CT images with low doses (40–100 mAs) from a CT image with a high dose (120 mAs). In contrast, our simulated results showed that the error of the SD for a Catphan phantom was about 3 % and that for a TOS phantom was less than 1 %. Therefore, our simulated results have an equivalent accuracy in comparison with those of previous studies.

Our method has several limitations. First, it is not certain that our method can be applied to a different scan type or rotation time because our study was performed only for the acquisition parameters shown in Table 1. We need to investigate the effects of different acquisition parameters in the future. Second, our method would not be applicable to ultra-low-dose simulation in which the effect of electrical noise is increased dominantly, because our noise model

was formulated with an overall term including both electric and quantum noise. Third, our proposed method could not be applied to CT images reconstructed by the iterative reconstruction (IR) algorithm because, instead of the reconstruction filter in FBP, a cost function for iterative computation is necessary for the low-dose simulation, which is beyond the scope of this study [12]. A low-dose simulation from CT images obtained by the IR algorithm is an important future task.

As shown in Fig. 9, the noise texture of the simulated low-dose CT image obtained by our proposed method seems to be comparable to that of the real low-dose CT image, whereas the slice positioning of these images differed somewhat on different days of the examinations. Because we do not have sufficient clinical databases to evaluate image quality further, a detailed evaluation of our proposed method for clinical CT images should be performed so that an optimal exposure dose can be determined in the near future.

5 Conclusion

We have developed a computerized scheme for producing simulated low-dose CT images from real high-dose images without use of the raw sinogram data. The noise model and the reconstruction filter were determined from access-friendly phantoms such as an ACR phantom. The simulated low-dose CT images produced by our proposed scheme could be used for determining optimal dose settings of various clinical CT examinations clinically without time-consuming experiments and risks in terms of increasing patient dose.

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Compliance with ethical standards

Conflict of interest The authors declare that they have no conflict of interest.

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