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Effect of Scatter Radiation to Image Noise in Cone Beam CT

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INTRODUCTION

Since the pioneer work of the Dynamic Spatial Reconstructor (DSR) in early 80's, cone beam CT has been studied by several groups (1, 2, 3, 4). They demonstrated isotropic resolutions in 3-dimensional imaging of large volumes, and suggested possibilities of 4-dimensional imaging (dynamic volume imaging) because cone beam CT acquire a large volume data with one rotation of X-ray tube-detector pair. However, their efforts have not resulted in routine clinical CT except imaging of high-contrast objects such as lung or blood vessels enhanced by contrast agent. It was because all groups failed to demonstrate the same image quality for low contrast objects as conventional CT due to narrow dynamic range of X-ray TV system employed as 2-dimensional detectors.

Recently a so-called flat-panel imager has been introduced to clinical radiology. It is a digital X-ray imaging device, based on the solid-state technologies developed for active matrix, flat-panel displays, and is capable of high-sensitive and high-speed imaging with wide dynamic range (5). It is expected that cone-beam CT could produce images of low contrast objects comparable of conventional CT if the solid-state detector such as flat-panel imager be employed in stead of X-ray TV system because of wider dynamic range. The other candidate of solid-state detector is an extension of multiple-row detector array equipped in multi-slice CT (6).

Another potential drawback with cone-beam CT is larger amount of scattered X-rays. These X-rays may enhance the noise in reconstructed images, and thus affect low contrast detectablity. The aim of this work was to estimate scatter fractions and effects of scatter to image noise, and was to seek methods of improving the image quality in cone beam CT with solid-state detectors.

MATERIALS AND METHODS

Image Noise Formulation

The noise in reconstructed image has a relationship to the noise in X-ray intensity measurements. The noise-to-signal ratio in reconstructed image is defined as σ^2/μ^2, where μ is the true value of attenuation coefficient at a point and σ is the variance in a set of measurements of the attenuation at that point. Chesler et al. proposed a formula of σ^2/μ^2 for a convolution algorithms in parallel beam geometry (7). Because Feldkamp algorithm (8), which is typically used in cone-beam reconstruction, is an extension of convolution algorithm, Chesler's formula is a good approximation of noise-to-signal ratio of reconstructed image in cone-beam CT if cone-angle is not a quite large. It follows as,

$$\frac{\sigma^2}{\mu^2} = \frac{\pi^2\sigma_p^2 a}{\mu^2 n}\int_{-\infty}^{\infty}|G(k)|^2\,dk \qquad (1)$$

where n is the number of views, a is the linear sampling distance and σ_P is the standard deviation of projections, that is,

$$\sigma_p = \left| \frac{d}{dI}(\ln \frac{I_0}{I})\sigma_I \right| = \frac{\sigma_I}{I} \qquad (2)$$

where I_0 and I are incident and transmitted X-ray intensities, respectively, and σ_I is the standard deviation of transmitted X-ray intensity. Therefore σ_P is the noise to signal ratio of X-ray intensity measurements. *G(k)* is the correction function in k-space (Fourier domain). For the Shepp and Logan correction function (9),

$$G(k) = \begin{cases} \frac{1}{\pi a}|\sin(\pi k a)| & (|k| \le 1/(2a)) \\ 0 & (|k| \ge 1/(2a)) \end{cases} \qquad (3)$$

If we substitutes Eqs.(2) and (3) to Eq.(1),

$$\left(\frac{\sigma}{\mu}\right)^2 = K^2\left(\frac{\sigma_I}{I}\right)^2 \qquad (4)$$

where

$$K^2 = \frac{1}{2\mu^2 a^2 n} \qquad (5)$$

Noise in Intensity Measurement

The noise to signal ratio of X-ray intensity measurements is approximately given by,

$$\frac{\sigma_I}{I} = \frac{(N_P + N_S + N_D^2)^{1/2}}{N_P} \qquad (6)$$

where *Np* and *Ns* are the numbers of primary and scattered X-ray photons absorbed in a detecror element per view, respectively. N_D is the addititve electronic noise in the detector including read-out electronics, and is converted to X-ray photon numbers because the quantum sink exists at the stage of X-ray absorption.

Estimation of Scatter

Scatter to primary ratio was estimated by a Monte Carlo simulation. The geometry of cone beam system is given in Fig.1. The patient was modeled as a water cylinder of diameter D (D=200 or 300mm) and of infinite length. The size of irradiation field was 600mm in transverse direction at the detector. It was varied in axial direction and was denoted by L. In the simulation X-ray tube voltage was 120 kVp and a compensating bow-tie filter was inserted between the source and patient.

RESULTS AND DISCUSSION

Fig. 2 shows the scatter to primary ratio at the center of midplane, estimated by the simulation. We estimated the noise to signal ratio in the reconstructed image from this ratio, using the values of parameters listed in Table 1. The size of detector element was assumed 1.0x1.0 mm^2, and X-ray exposure was assumed 1.0 mAs per view. These assumption resulted 1.1×10^4 or 1.6×10^3 primary X-ray photons absorbed in the detector element for D=200 or D=300, respectively, at the center of midplane.

Table 2 shows the examples of estimation results. The condition chosen was L=400mm, n=360 and X-ray exposure of 1.0 mAs per view, which corresponded to exposure dose 2.5-3.0 times larger than conventional CT (10). The noise to signal ratios in the X-ray intensity measurement σ_I/I were shown separately in their components (primary, scatter and detector). These results show substantial contribution of scatter radiation to image noise in cone-beam CT, and detector noise is also substantial for the cylinder of D=300mm. Scatter rejection by a collimator is possible to decrease the image noise, and is now under study.

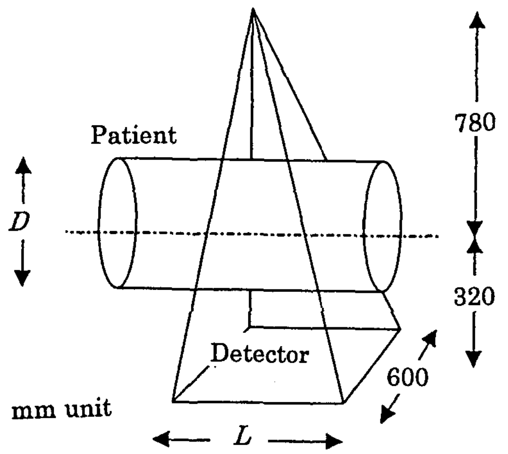

Fig.1 Geometry of cone beam CT

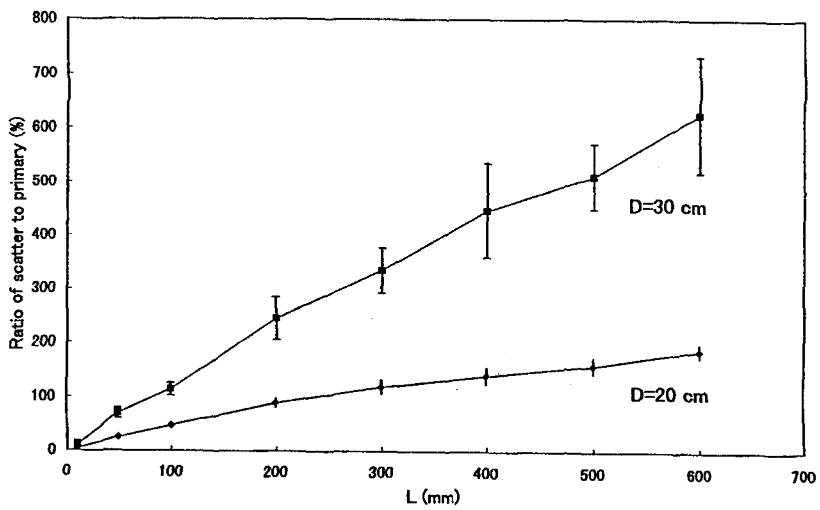

Fig.2 Scatter to primary ratio vs. axial length of irradiation fields L

Table 1. Parameter list used in this study

μ_0	0.20 cm^{-1}
a	1.0 mm
n	360
K	1.85
N_p	1.1x10^4 (*D*=200)
	1.6x10^3 (*D*=300)
L	400 mm
Scatter to primary ratio	1 (*D*=200)
	5 (*D*=300)
N_D	50

Table 2. Estimated noise to signal ratios

D	σ_I/I				σ/μ
	primary	scatter	detector	total	
200mm	0.95 %	0.95	0.45	1.42	2.8
300	2.50	5.59	3.12	6.87	12.7

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