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Attenuation correction using combination of a parallel hole collimator and an uncollimated non-uniform line array source

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Attenuation correction is very important for quantitative SPECT imaging. We designed an uncollimated non-uniform line array source (non-uniform LAS) for attenuation correction based on transmission computed tomography (TCT) using Tc-99m and compared its performance with an uncollimated uniform line array source (uniform LAS) in a thorax phantom study. This non-uniform LAS was attached to one camera head of a dual-head gamma camera, and transmission data were acquired with another camera head with a low-energy, general purpose, parallel-hole collimator at 50 cm-distance apart from the source. The modified TEW using a subtraction factor of 1.0 was employed to correct scattered Tc-99m photons for transmission data. In the phantom experiment, eight TCT data were acquired with the scanning time changed from 2 minutes to 20 minutes for each LAS. The Tc-99m attenuation coefficient (μ) maps with the non-uniform LAS and uniform LAS improved the statistical count variation in the mediastinum filled with water as the scanning time got longer. The Tc-99m μ-map with the non-uniform LAS and 6 minutes of scanning time had equal quality at the center of the thorax phantom to that with the uniform LAS and 16 minutes of scanning time. In conclusion, for the TCT imaging with combination of the parallel hole collimator and uncollimated Tc-99m external source the non-uniform LAS can reduce the Tc-99m radioactivity or the TCT scanning time compared with the uniform LAS.

Key words: attenuation correction, uncollimated non-uniform line array source, parallel hole collimator, myocardial phantom SPECT, Tc-99m

INTRODUCTION

IN SINGLE PHOTON EMISSION COMPUTED TOMOGRAPHY (SPECT) imaging, transmission computed tomography (TCT) data for attenuation correction is very useful to obtain quantitative SPECT image counts, particularly for non-uniform media. Many TCT methods using various external source configurations have been investigated for attenuation correction in SPECT imaging.[1–3] A TCT imaging method using the combination of an uncollimated uniform line array source (uniform LAS) and a parallel-hole collimator to obtain accurate attenuation coefficient (μ) maps was previously proposed by two of the present coauthors.[4,5] This TCT method makes it simple to obtain the μ-map and does not require major camera modification. As the uncollimated external source generates many scatter events in the transmission data, we employed the modified triple energy window (TEW) technique to perform accurate scatter correction for those transmission data.[6] However, it was found that this modified TEW method based on the scatter counts-subtraction technique causes poor

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statistical transmission counts in the center of the body. In this study we propose the use of an uncollimated non-uniform line array source (non-uniform LAS) instead of the uniform LAS to obtain better μ-maps for the TCT method using the combination of the LAS and the parallel-hole collimator.

MATERIALS AND METHODS

All data acquisitions were performed with a dual-headed gamma camera (GCA-7200A/DI, Toshiba, Japan) and a nuclear medicine computer system (GMS-5500/DI). Each camera was equipped with a low-energy, general purpose (LEGP), parallel-hole collimator. As uncollimated transmission sources, an uncollimated uniform line array source with Tc-99m solutions of 1.11 GBq (uniform LAS, 55 cm × 40 cm, 50 m*l*) and an uncollimated non-uniform line array source including Tc-99m (140 keV) solutions of 666 MBq (non-uniform LAS, 50 cm × 25 cm, 18 m*l*) were used. The uniform LAS had a 5 mm interval,[4,5] and the non-uniform LAS was designed to make the density of the line source high at the center and low at both sides while changing the interval of the line group every 5 mm (Fig. 1). Each LAS was made of a fluoroplastic tube with an internal diameter of 1 mm and housed in a 10 mm-thick acrylic plate. For TCT scatter correction at 140 keV-photopeak, triple energy window (TEW) mode (20% main energy window at 140 keV, 7% lower sub-energy window and 0% upper sub-energy window) was considered. A thorax phantom (Type RH-2, Kyoto Kagaku, Japan) with a mediastinum, two lungs, and a spinal cord was used in this study. The heart phantom with a myocardium and a chamber was placed in the mediastinum.

For the thorax phantom with the myocardial phantom, TCT and SPECT data were acquired with a continuous rotation mode in 60 views over 360° and in 128 × 128 matrices with a pixel size of 4.3 mm.

A. TCT imaging for the thorax phantom

The mediastinum of the thorax phantom was filled with non-radioactive water, and the heart phantom was not put in the mediastinum (Fig. 2). For both the non-uniform LAS and uniform LAS, TCT data acquisition time was varied from 2 minutes to 20 minutes in 2-minute increments. The distance between the non-uniform LAS or uniform LAS and the LEGP collimator was 62 cm. Following scatter correction by the modified triple energy window (TEW) method and two-dimensional Butterworth filtering (order of 8 and cutoff frequency of 0.14 cycles/pixel), TCT images with 1 pixel thickness were reconstructed by a filtered backprojection with a ramp filter.

B. SPECT imaging for Tl-201 myocardial phantom

The heart phantom with Tl-201 of 10 MBq was inserted in the mediastinum of the thorax phantom as shown in Figure 2. The Tl-201 radioactivity in the lungs and mediastinum was 18 MBq and 37 MBq, respectively. For Tl-201 imaging with the TEW acquisition, a 47% main energy window was set at 73 keV, and two 7% sub-energy windows were set at both sides of the main energy window. The projection data acquisition time was 20 minutes. The radius of rotation was 25 cm. After the conventional TEW scatter correction and Butterworth filtering (order of 8 and cutoff frequency of 0.14 cycles/pixel), SPECT images were reconstructed by a filtered backprojection with a ramp filter. The Chang iteration method (three iteration times) and the attenuation maps by the non-uniform LAS were used for attenuation correction of SPECT data.

C. Triple energy window (TEW) method for scatter correction

The primary counts C_p within the main energy window are estimated by the following equation:

$$C_p = C_m - C_s \tag{1}$$

where C_m is the total counts within the main energy window and C_s is the estimated scatter counts. The scatter counts C_s is calculated by

$$C_s = K \cdot (C_l \cdot M_w/S_l + C_u \cdot M_w/S_u) \tag{2}$$

where C_l and C_u are the counts within the lower and upper

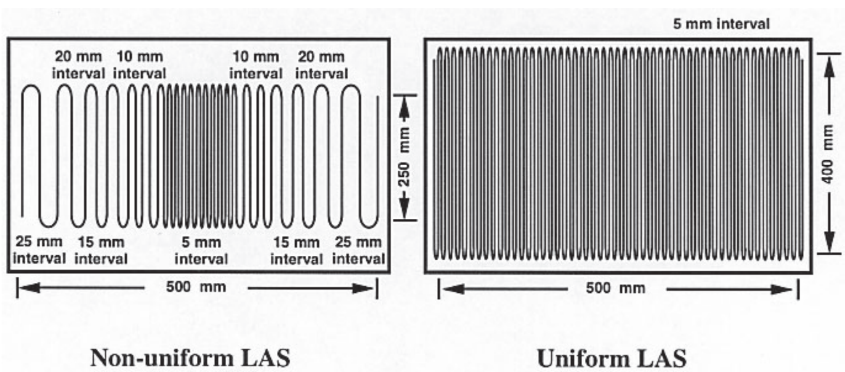


Fig. 1 Uncollimated Tc-99m line array source configurations.

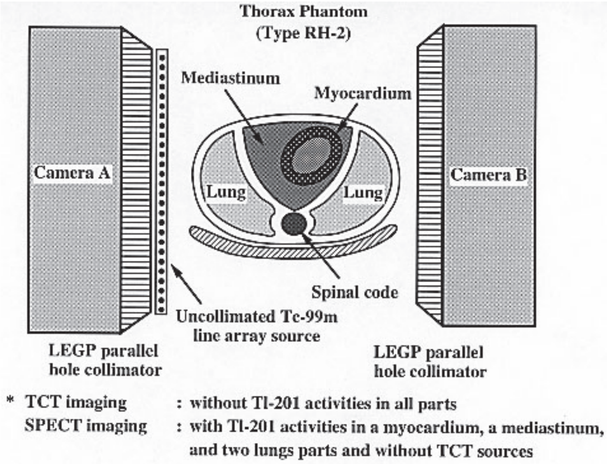


Fig. 2 TCT and SPECT imaging using a thorax phantom.

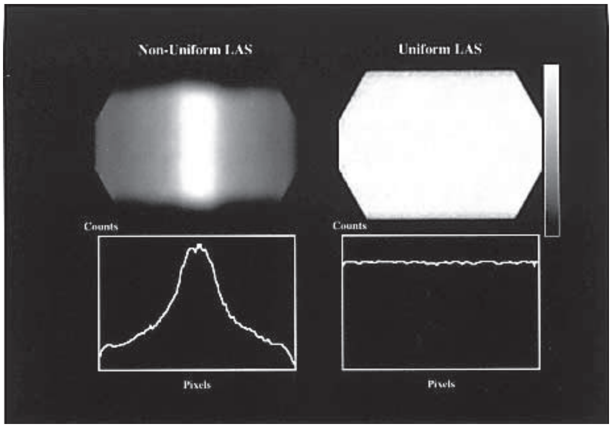
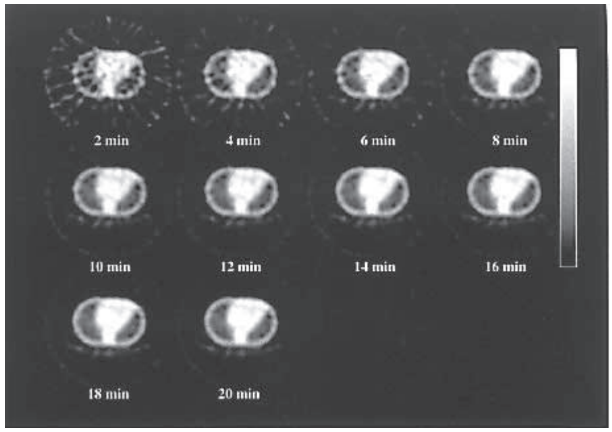
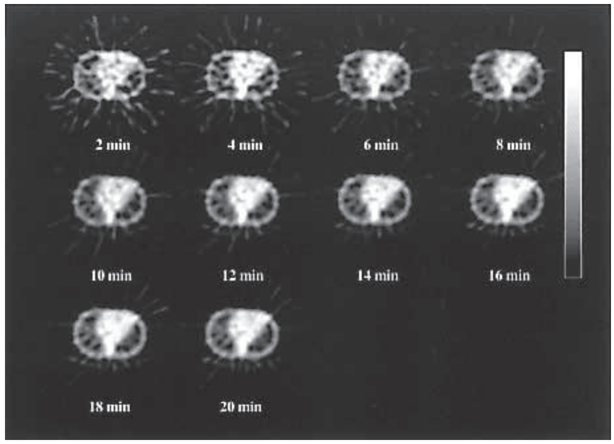


Fig. 3 Comparison between the non-uniform LAS and the uniform LAS on planar images (blank data) and count profile curves. Due to the blurred spatial resolution, the blank image with the uniform LAS had a uniform count distribution at the whole field of view, but that with the non-uniform LAS gave a smooth mountain shape with high count rates at the center and low count rates at both sides.



A



B

Fig. 4 Comparison between the non-uniform LAS (A) and the uniform LAS (B) on TCT attenuation maps. The 6 minutes-TCT scanning time image with the non-uniform LAS had almost equal quality at the center of the thorax phantom (mediastinum) to the 16 minutes-TCT scanning time image with the uniform LAS.

sub-energy windows, respectively. M_w, S_l and S_u are the widths of the main energy window, lower and upper sub-energy windows, respectively. K is the subtraction factor. The K value of 0.5 (the conventional TEW method) is employed for SPECT data[7,8] and the K value of 1.0 (the modified TEW method) for TCT data.[6]

RESULTS

A. Non-uniform LAS and uniform LAS images

Planar images (blank data) and count profile curves for the non-uniform LAS and uniform LAS are shown in Figure 3. Each count profile curve was obtained by the integration of 10 pixel-horizontal lines including the center line of the image. Due to the blurred spatial resolution, the planar blank image with the uniform LAS had a uniform count distribution at the whole field of view. However, the blank image with the non-uniform LAS gave a smooth mountain-shape with high count rates at the center and low count rates at both sides, with the ratio for these count rates being 1 : 0.2.

B. TCT images using the non-uniform LAS and uniform LAS

The comparison of TCT images using the non-uniform LAS and uniform LAS are shown in Figure 4. The TCT images with the non-uniform LAS and uniform LAS improved statistical count variation in the mediastinum filled with water as the scanning time became longer. However, the 6 minutes-scanning time TCT image with the non-uniform LAS had almost equal quality at the center of the thorax phantom (mediastinum) to that of the

Table 1 Mean attenuation coefficient and standard deviation (SD) in TCT imaging with the non-uniform LAS and uniform LAS

TCT scanning time (minutes)	Attenuation coefficient (cm^{-1})			
	Non-uniform		Uniform	
	mean	SD	mean	SD
2	0.158	0.023	0.153	0.028
4	0.162	0.016	0.170	0.030
6	0.158	0.016	0.166	0.025
8	0.158	0.017	0.169	0.016
10	0.156	0.016	0.167	0.018
12	0.154	0.014	0.165	0.018
14	0.153	0.012	0.165	0.016
16	0.153	0.012	0.165	0.016
18	0.152	0.013	0.166	0.015
20	0.151	0.012	0.164	0.015

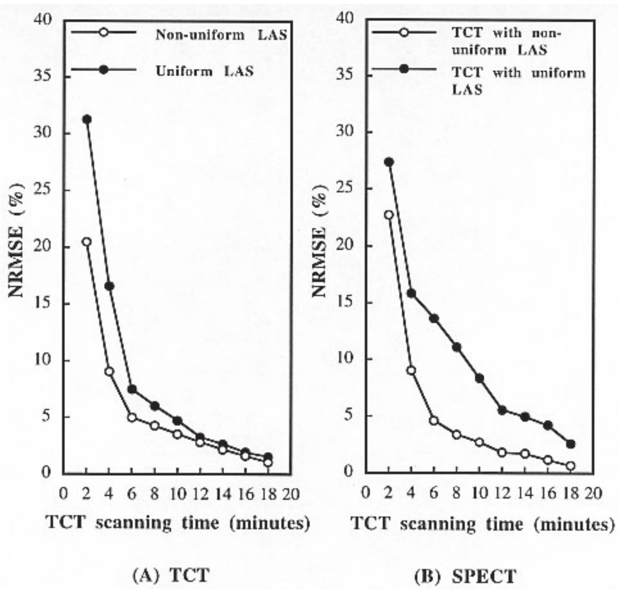
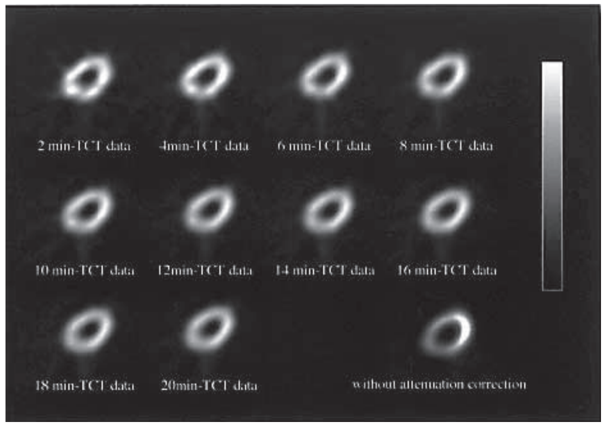


Fig. 5 Comparison of normalized root mean square error (NRMSE) between the non-uniform LAS and the uniform LAS on TCT images (A) and SPECT images (B) with different scanning times. By means of the non-uniform LAS, an acceptable NRMSE of less than 5% in the SPECT image was achieved with the 6 minutes-TCT scanning time.



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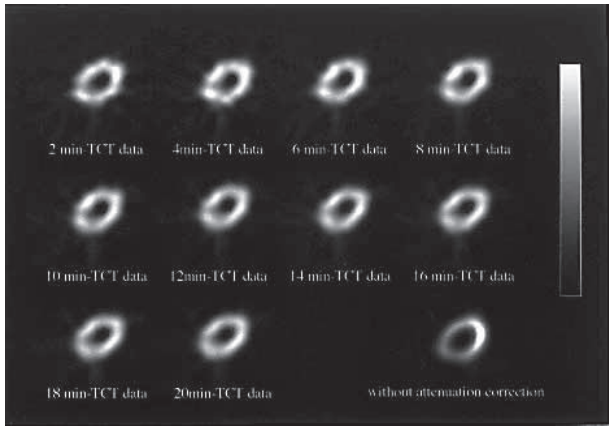

B

Fig. 6 Tl-201 myocardial phantom SPECT images (transaxial) reconstructed by attenuation maps with different TCT scanning times using the non-uniform LAS (A) and uniform LAS (B). At the shorter TCT scanning time, the attenuation corrected SPECT images with the non-uniform LAS had less noise than those with the uniform LAS.

16 minutes-scanning time TCT image with the uniform LAS. We also analyzed the attenuation coefficient value and statistical variation, and normalized root mean square error (NRMSE) in an ROI set on the mediastinum for all the TCT images. Each TCT image with 20 minutes-scanning time (thought as the best data) was used as the reference data in the normalization for root mean square errors in TCT images with various scanning times. Table 1 lists the mean attenuation coefficient values and standard deviations (SD) for the non-uniform LAS and uniform LAS. The non-uniform LAS data had attenuation coefficient values closer to the theoretical value (0.154 cm^{-1}) and smaller SDs in comparison with the uniform LAS data. Furthermore, even 6 minutes scanning time TCT image gave an acceptable NRMSE of less than 5% as illustrated in Figure 5A.

C. SPECT images corrected for attenuation using the non-uniform LAS

Tl-201 myocardial SPECT images reconstructed with the attenuation correction using the non-uniform LAS and the uniform LAS are shown in Figure 6. The TCT data with shorter scanning time made attenuation corrected SPECT images noisier. The NRMSE for the SPECT images was also investigated. The SPECT image corrected for attenuation by the 6 minutes scanning time TCT data gave acceptable quantitation in less than 5% NRMSE (Fig. 5B).

DISCUSSION

To perform accurate attenuation correction, a number of TCT imaging techniques have been proposed.[1] In these techniques, there are three types of TCT imaging method for the use of a parallel-hole collimator: uncollimated flood source type, collimated flood source type and scanning line source type.[2–6,9–11] The scanning line source technique requires complicated electronic equipment, but the flood source techniques are easily applicable to general gamma camera SPECT systems. Regarding the flood source methods, since the uncollimation of the source results in a map of broad attenuation coefficients, the collimated flood source is practical and preferable. However, the collimated flood source technique requires an extra source collimator and a large amount of radioactivity.

Recently, a TCT imaging technique using the combination of an uncollimated uniform line array source (uniform LAS) and a parallel-hole collimator was developed for the dual-head gamma camera SPECT system.[4,5] Although this uncollimated uniform LAS technique is similar to the method reported by Celler et al.[12] with respect to the use of the multiple line source array, it does not require much radioactivity compared with the collimated flood source. If such an uncollimated transmission source is employed for TCT imaging, accurate scatter correction is needed to obtain quantitative SPECT values.[13] The triple energy window (TEW) method is a simple one to correct for scatter and generally uses the subtraction factor K of 0.5 (Eq. 2) for ECT imaging. However, we found that the conventional TEW method with $K = 0.5$ was not sufficient to compensate for scatter included in transmission data using the uncollimated uniform LAS and proposed the use of the $K = 1.0$ for TCT imaging (the modified TEW method).[6] Although the ratio of the scatter and primary counts (scatter fraction) varies pixel by pixel for various source configurations, we used the mean K value of 1.0 for the non-uniform LAS. The determination of a more suitable K value pixel by pixel is a focus for future research. Since the modified TEW method subtracts more scatter counts from projection images, the count rate in the center of the body decreases and statistical variation increases. In this work, to increase the counting rate at the center of the body without increasing the total amount of flood source activity, we designed an uncollimated non-uniform line array source (non-uniform LAS) which has a non-uniform count distribution (high count rates at the center and low count rates at both sides).

The non-uniform LAS consisted of a line source folded up in a thin acrylic plate, with the pitch of this winding line only 25 mm at both sides of the source. The spatial resolution changes according to the type of parallel-hole collimator and the distance between the imaging camera head and the source. For the non-uniform LAS designed in this study, relatively smoothed blank data were obtained because of the very poor spatial resolution (~42 mm-FWHM) at the source-to-imaging camera distance of 62 cm (Fig. 3). Therefore, we employ the LEGP collimator and the fixed distance of 62 cm between that collimator and the source for TCT imaging with the non-uniform LAS.

In clinical use, we must consider when TCT data should be acquired on the SPECT imaging study. Usually, TCT and ECT data sets are obtained either simultaneously or sequentially. The simultaneous acquisition is preferable both to reduce imaging time and to avoid misregistration of the TCT and ECT data sets due to patient movement between the studies.[11,14] However, since the simultaneous acquisition results in cross-contamination of the TCT and ECT data because of overlapping of different isotopes' photon energies, the sequential imaging protocol may be recommended.[15] In such a sequential imaging protocol, shortening of the TCT scan time is highly desirable to reduce the total imaging time and exposure to the patient. In this study, we examined the non-uniform LAS in the thorax phantom experiment and compared its performance with the uniform LAS on the assumption that the uncollimated flood source might be used in clinical practice. The results of the phantom experiment showed that even 6 minutes TCT scan data with the non-uniform LAS gave acceptable attenuation coefficient values and produced quantitative SPECT images, with these data almost equal to the 16 minutes TCT scan data with the uniform LAS in the center of the body phantom. Therefore, the TCT scan time can be reduced with the non-uniform LAS in comparison with the uniform LAS in the sequential TCT and ECT imaging.

Although it has been found that the non-uniform LAS has greater advantages than the uniform LAS, it is necessary to investigate the usefulness of the non-uniform LAS for a large spectrum of patient sizes in both men and women. Furthermore, as the amount of source activity is a trade-off between the image quality and scan time, the optimum amount of source activity must be investigated when using the non-uniform LAS.

CONCLUSIONS

The TCT imaging with the combination of a parallel hole collimator and an uncollimated non-uniform LAS is a simple method for attenuation correction. This proposed TCT imaging technique can give more accurate attenuation coefficient maps with less external radioactivity than the uncollimated uniform LAS.

REFERENCES

1. King MA, Tsui BMW, Pan TS. Attenuation compensation for cardiac single-photon emission computed tomographic imaging: part 1. impact of attenuation and methods of

estimating attenuation maps. *J Nucl Cardiol* 1995; 2: 513–524.

2. Ogawa K, Takagi Y, Kubo A, Hashimoto S, Sannmiya T, Okano Y, et al. An attenuation correction method of single photon emission computed tomography using gamma ray transmission CT. *KAKU IGAKU (Jpn J Nucl Med)* 1985; 22: 477–490.
3. Ogawa K, Kubo A, Hashimoto S, Morozumi T, Nakajima M, Yuta S, et al. An attenuation correction of SPECT image using transmission data acquired with dual head gamma camera system. *Med Imaging Technol* 1985; 3S: 103–104.
4. Hashimoto J, Sannmiya T, Ogasawara K, Kubo A, Ogawa K, Ichihara T, et al. Scatter and attenuation correction for quantitative myocardial SPECT. *KAKU IGAKU (Jpn J Nucl Med)* 1996; 33: 1015–1019.
5. Ichihara T, Maeda H, Yamakado K, Motomura N, Matsumura K, Takeda K, et al. Quantitative analysis of scatter- and attenuation-compensated dynamic single-photon emission tomography for functional hepatic imaging with a receptor-binding radiopharmaceutical. *Eur J Nucl Med* 1997; 24: 59–67.
6. Kojima A, Matsumoto M, Tomiguchi S, Katsuda N, Yamashita Y, Motomura N. Accurate scatter correction for transmission computed tomography using an uncollimated line array source. *Ann Nucl Med* 2004; 18: 45–50.
7. Ogawa K, Harata Y, Ichihara T, Kubo A, Hashimoto S. A practical method for position-dependent compton-scatter correction in single photon emission CT. *IEEE Trans Med Imag* 1991; 10: 408–412.
8. Ichihara T, Ogawa K, Motomura N, Kubo A, Hashimoto S. Compton scatter compensation using the triple-energy window method for single- and dual-isotope SPECT. *J Nucl Med* 1993; 34: 2216–2221.
9. Bailey DL, Hutton BF, Walker PJ. Improved SPECT using simultaneous emission and transmission tomography. *J Nucl Med* 1987; 28: 844–851.
10. Tsui BMW, Gullberg GT, Edgerton ER, Ballard JG, Perry JR, McCartney WH, et al. Correction of nonuniform attenuation in cardiac SPECT imaging. *J Nucl Med* 1989; 30: 497–507.
11. Tan P, Bailey DL, Meikle SR, Eberl S, Fulton RR, Hutton BF. A scanning line source for simultaneous emission and transmission measurements in SPECT. *J Nucl Med* 1993; 34: 1752–1760.
12. Celler A, Sitek A, Stoub E, Harrop P, Lyster D. Multiple line source array for SPECT transmission scans: simulation, phantom and patient study. *J Nucl Med* 1998; 39: 2183–2189.
13. Ogawa K, Kubo A, Ichihara T. Estimation of scattered photons in gamma ray transmission CT using Monte Carlo simulations. *IEEE Trans Nucl Sci* 1997; 44: 1225–1230,
14. Frey EC, Tsui BMW, Perry JR. Simultaneous acquisition of emission and transmission data for improved thallium-201 cardiac SPECT imaging using a technetium-99m transmission source. *J Nucl Med* 1992; 33: 2238–2245.
15. Tomiguchi S, Kira T, Kojima A, Matsumoto M, Nishi J, Katsuda N, et al. Attenuation correction in Tl-201/Tc-99m myocardial SPECT. *Eur J Nucl Med* 1999; 26: 1186.