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Development of a collimator blurring compensation method using fine angular sampling projection data in SPECT

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Due to the collimator aperture, spatial resolution of SPECT data varies with source-to-detector distance. Since the radius of detector rotation is bigger when scanning larger patients, spatial resolution is degraded in these cases. Emitted gamma rays travel not only along the central axis of the collimator hole but also off-axis due to the collimator aperture. However, an off-axis ray at one angle would be a central-axis ray at another angle; therefore, raw projection data at one angle can be thought of as an ensemble of central-axis rays collected from a small arc equal to the collimator aperture. Thus, fine angular sampling can compensate for collimator blurring. By using a sampling pitch of less than half the collimator aperture angle, compensation was performed by subtracting the weighted sum of the projection data from the raw projection data. Collimator geometry and detector rotation radius determined the weighting function. Cylindrical phantom with four different-sized rods and torso phantom for Tl-201 cardiac SPECT simulation were used for evaluation. Aperture angle of the collimator was 7 degrees. Projection sampling pitch was 2 degrees. In both phantom studies, the proposed method showed improvement in contrast and reduction of partial volume effect, thereby indicating that the proposed method can compensate adequately for image blurring caused by the collimator aperture.

Key words: SPECT, collimator aperture, spatial resolution, filter, blurring

INTRODUCTION

IN SINGLE PHOTON EMISSION COMPUTED TOMOGRAPHY (SPECT), spatial resolution is the most important factor in determining image quantification and image contrast. Especially in clinical studies, detection sensitivity for tumor lesions and defect areas greatly depends on spatial resolution. SPECT spatial resolution is determined by the intrinsic resolution of the detector and blurring due to the collimator aperture. The intrinsic resolution of the detector can be significantly improved by using a pixel detector such as a semiconductor detector or a pixellated scintilator.[1] Blurring due to the collimator aperture can be improved by adjusting the collimator geometry (greater length or smaller hole), coupled with reduced sensitivity. Another approach is to use data processing to compensate for collimator blurring. Several methods have been investigated. These can be roughly divided into three categories, i.e. the filtering method, frequency distance relationship (FDR) method and iterative method. The filtering method uses a high frequency component enhanced filter (such as a Wiener filter).[2–6] Processing is fast and simple, but image distortion occurs. The FDR method can be processed analytically, is significantly affected by statistical noise.[7,8] The iterative method is little affected by statistical noise and can process precise compensation.[9–13] However, the iterative method requires a lengthy processing time. Therefore, there is no practical method of collimator blurring compensation, which can be used in routine clinical study.

In this work, we have evaluated a collimator blurring reduction method that uses a sampling pitch of less than

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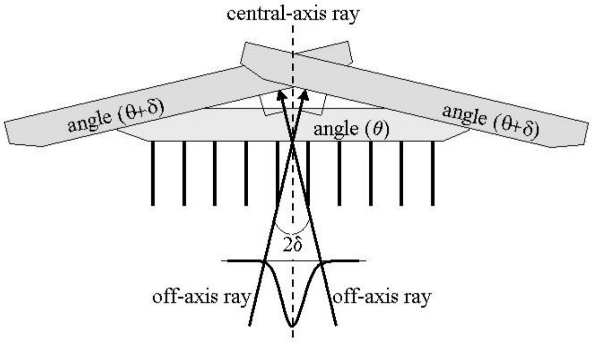


Fig. 1 The relationship between central-axis ray and off-axis ray using the proposed collimator blurring reduction method.

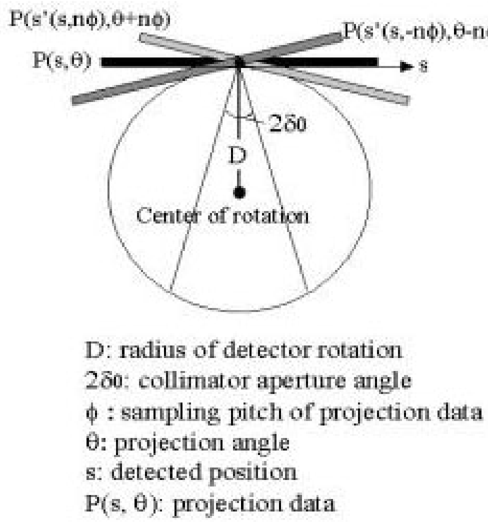
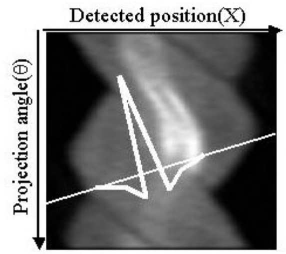


Fig. 2 Illustration of the collimator blurring reduction method using fine angular sampling projection data and weighting function (reverse filter) in SPECT.



Detected position (X)

Projection angle(θ)									
	0.000	0.000	0.000	0.000	0.000	0.000	-0.125	-0.125	0.000
	0.000	0.000	0.000	0.000	0.000	1.500	0.000	0.000	0.000
	0.000	-0.125	-0.125	0.000	0.000	0.000	0.000	0.000	0.000

Fig. 3 The shape of the used weight function is shown, which has two dimensions consisting of projection angle and detected position. This weight function is subjected to sinogram data.

half the collimator aperture angle and the 2D filtering to sinogram data. The proposed method uses the same assumption as Ogawa et al.[11] This assumption is that the measured projection data are a weighted sum of several neighboring projection data. Two phantom tests were performed to evaluate whether the proposed method can compensate for the poor spatial resolution SPECT data which are acquired with a large detector rotation radius.

MATERIALS AND METHODS

A. Theory

Emitted gamma rays travel along the central axis of the collimator hole as well as the off-axis due to the collimator aperture. Off-axis rays can be regarded as central-axis rays from another projection angle; therefore, raw data can be thought of as an ensemble of central-axis rays from within an arc equal to that created by the collimator aperture (Fig. 1). If only central-axis rays could be acquired, these data could be used to compensate for collimator blurring by subtracting the projection data from the central-axis set within the collimator aperture arc from the actual projection data. Using this principle, collimator blurring was compensated for by using fine angular sampling. The appropriate sampling pitch was less than half the collimator aperture angle. To compensate projection data at θ degrees, neighboring projection data at $\theta \pm \delta$ degrees were used. "δ" represents half the collimator aperture angle. The weighting function was determined by the collimator geometry and the detector rotation radius, assuming that collimator blurring is triangle-shaped. The sum of weighting values should be normalized to 1.0 for quantitation. Compensation was performed by subtracting the weighted sum of the projection data from the measured projection data (Fig. 2). The measured projection data $P(s,q)$ are compensated for with the following equation.

Compensated $P(s,\theta)$

$$= P(s,\theta) - \sum_{n=-m}^{n=+m} w(s'(s,n\phi),n\phi) * \{P(s'(s,n\phi),\theta + n\phi)\} \quad (1)$$

where

$m\phi < \delta_0$ (2 δ_0 is collimator aperture angle)
ϕ is sampling pitch of projection data.
$w(s'(s,n\phi),n\phi)$ is the weighting function.
$P(s'(s,n\phi),\theta + n\phi)$ is assumed to be coincident with the off-axis ray of $P(s,\theta)$ due to collimator aperture $n\phi$.
A symbol "*" represents a convolution process.

B. Acquisition and Processing

We conducted a feasibility study of the proposed method using phantoms. All images in this investigation were acquired with a GCA-7200A/DI dual-head gamma camera system (Toshiba, Tokyo) equipped with a low-energy general purpose (LEGP) parallel hole collimator with an aperture angle of 7 degrees. Acquired data were processed with a GMS-5500A/PI medical image processor (Toshiba, Tokyo). Emission projection data were acquired using a

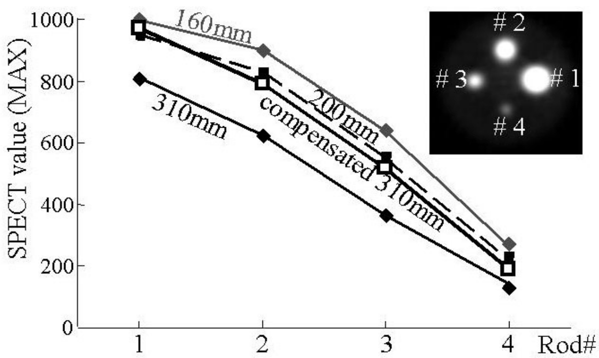


Fig. 4 The result of the phantom study 1. This graph shows the maximum SPECT value of each rod acquired with 160 mm, 200 mm and 310 mm. The data acquired with a 300 mm is compensated using the proposed method.

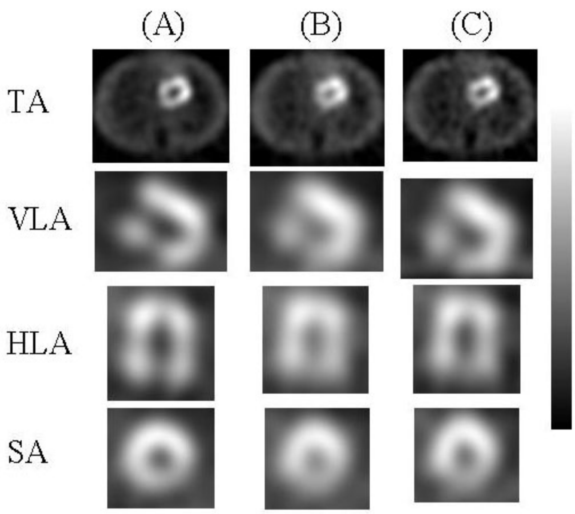


Fig. 5 The result of the phantom study 2. Three kinds of data set: automatic body contouring (A), detector rotation radius of 310 mm (B) and compensated detector rotation radius of 310 mm (C) are shown. Transverse images (TA), vertical long-axis images (VLA), horizontal long-axis images (HLA) and short-axis images (SA) are displayed.

continuous rotation scan mode with a 128 × 128 matrix (4.3 mm/pixel). Because the sampling pitch of the projection angle must be less than half the collimator aperture angle, 2 degrees/view (180 views/rotation) was applied. The energy window was set over the photopeak of the used radioisotope (Tc-99m or Tl-201) with 20% of the photopeak energy width.

1) Phantom Study 1

A cylindrical water phantom (diameter of 20 cm, length of 20 cm) consisting of four rods was used. The diameters of the four rods were 10 mm, 20 mm, 30 mm and 40 mm. The four rods were filled with Tc-99m solution with the same specific radioactivity of 120 kBq/m*l*. The remaining area of the cylindrical phantom was filled with a 10 kBq/m*l* solution. Three SPECT scans were performed with three different detector rotation radii (160 mm, 200 mm and 310 mm). Acquisition time was 20 minutes/scan. The proposed method was applied only to the projection data acquired with a 310 mm detector rotation radius. The shape of the used weight function is shown in Figure 3. The parameter "m" in equation (1) was 1 (both neighboring project angles were used for this compensation). All projection data were subjected to 2D Butterworth filter processing (cutoff frequency of 0.33 cycle/cm and order 8) before reconstruction using a filtered backprojection method with a Shepp & Logan filter.

2) Phantom Study 2

The proposed method was also evaluated using a "Jaszczak" torso phantom (Dataspectrum, USA). Tl-201 was used to fill the cardiac insert (79 kBq/m*l*), liver (31 kBq/m*l*) and mediastinum (11 kBq/m*l*) to simulate 111 MBq Tl-201 myocardial SPECT. Two SPECT scans were performed with a detector rotation radius of 310 mm and automatic body contouring (detector at the minimum radius at each projection angle). In both scans, the acquisition times were 5, 10, 15 and 20 minutes/scan to evaluate the influence of statistical noise. The data processing conditions were the same as in phantom study 1. The reconstructed images were resliced to vertical long axis images, horizontal long axis images and short axis images. Coefficients of variation (C.V.: standard deviation divided by average value) were calculated to evaluate the influence of statistical noise due to scan time.

RESULTS

A. Phantom Study 1

According to the partial volume effect, the SPECT value of each rod decreases as the detector rotation radius increases or the rod diameter decreases. The maximum SPECT value of each rod was measured (Fig. 4). The maximum SPECT value of compensated data acquired with a 310 mm rotation radius was similar to that of the uncompensated data acquired with a 200 mm rotation radius. The results indicate that the proposed method can compensate for spatial resolution.

B. Phantom Study 2

Figure 5 shows three kinds of data set: automatic body contouring, detector rotation radius of 310 mm and compensated detector rotation radius of 310 mm. Transverse images, vertical long-axis images, horizontal long-axis images and short-axis images are displayed. In this study, improvement in contrast and reduction of the partial volume effect can be seen in the myocardial region. Figure 6 shows transverse images and C.V. acquired with scan times of 5, 10, 15 and 20 minutes. Because a uniform area should be used to calculate the C.V., the liver was set as the region of interest.

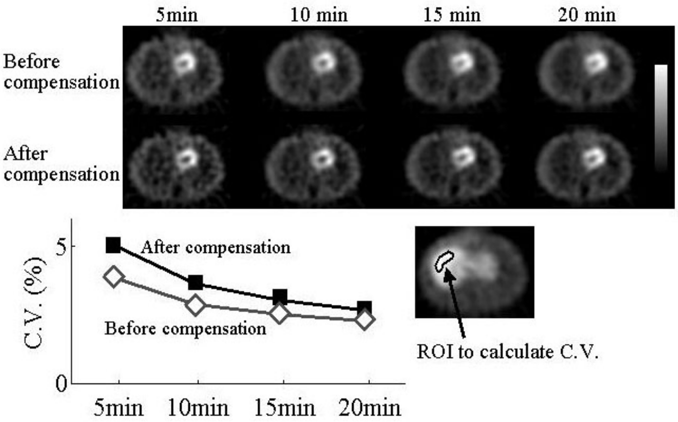


Fig. 6 The result of the phantom study 2. Transverse images both before and after compensation are shown with scan time 5, 10, 15 and 20 minutes. The C.V. values are also shown.

DISCUSSION

In our study of the collimator blurring reduction method, practicality has been considered to be the most important factor. Since the proposed method is categorized into a filtering method, processing is fast and simple. The method should be easy to use in clinical study and feature speedy data processing and adequate compensation performance. In this work, we evaluated the feasibility of the proposed method with a view to clinical use. Fine angular sampling increases the total acquisition time in step and shoot SPECT acquisition mode due to the detector moving time. However, when continuous rotation SPECT acquisition mode is applied, the total acquisition time does not increase in any angular sampling pitch. Because the added data processing is only 2D filtering to sinogram data, the proposed method can be quickly processed. Therefore, we believe that the proposed method can be applied to routine clinical studies and would be useful in a) large patient studies and b) scans without automatic body contouring (triple head or ring type gamma camera).

The proposed method can also be used to convert high-sensitivity data acquired with a high sensitivity (low resolution) collimator to high-resolution data. In this work, compensation of the spatial resolution was performed in the transverse plane (x and y directions) except for the axial (z) direction. If projection data tilted in the body axis is acquired, compensation of the axial (z) direction can be performed. However, although compensation of the axial (z) direction is theoretically possible, acquisition time increases by more than three times.

More investigation is required to find the optimal weight function and projection sampling pitch for the proposed method. Comparison with the FDR method and iterative method is needed. Finally, clinical study should be performed.

CONCLUSION

The feasibility of the proposed method has been evaluated using phantom studies. The results indicate that the proposed method can reduce collimator blurring and can be used in routine clinical study.

REFERENCES

1. Mori I, Takayama T, Motomura N. The CdTe detector module and its imaging performance. *Ann Nucl Med* 2001; 16: 487–494.
2. King MA, Doherty PW, Schwinger RB. A Wiener filter for nuclear medicine images. *Med Phys* 1983; 10: 876–880.
3. King MA, Doherty PW, Schwinger RB, Jacobs DA, Kidder RE, Miller TR. Fast count-dependent digital filtering of nuclear medicine images. *J Nucl Med* 1983; 24: 1039–1045.
4. Hon TC, Rangayyan RM, Hahn LJ, Kloiber R. Restoration of gamma camera-based nuclear medicine images. *IEEE Trans Med Imag* 1989; 8: 354–363.
5. Madsen MT, Park CH. Enhancement of SPECT images by Fourier filtering the projection image set. *J Nucl Med* 1985; 26: 395–402.
6. Honda N, Machida K, Tsukada J, Hosoba M. Optimal preprocessing Butterworth-Wiener filter for Tl-201 myocardial SPECT. *Eur J Nucl Med* 1987; 13: 404–407.
7. Edholm PR, Lewitt RM, Lindholm B. Novel properties of the Fourier decomposition of the sinogram. *Proceedings of the SPIE* 1986; 671: 8–18.
8. Lewitt RM, Edholm PR, Xia W. Fourier method for correction of depth-dependent collimator blurring. *Proceedings of the SPIE* 1989; 1092: 232–243.
9. Xia W, Lewitt RM. Iterative correction for spatial collimator blurring in SPECT. *Conference Record of the IEEE Nucl Sci Symposium* 1990; 2: 1158–1162.
10. Knesaurak K, King MA, Glick SJ, Penney BC. Investigation of causes of geometric distortion in 180 and 360 angular sampling in SPECT. *J Nucl Med* 1989; 30: 1666–1675.
11. Ogawa K, Katsu H. Iterative correction method for shift-variant blurring caused by collimator aperture in SPECT. *Ann Nucl Med* 1996; 10: 33–40.
12. Lau YH, Hutoon BF, Beekman FJ. Choice of collimator for cardiac SPECT when resolution compensation is included in iterative reconstruction. *Eur J Nucl Med* 2001; 28: 39–47.
13. Yokoi T, Shinohara H, Onishi H. Performance evaluation of OSEM reconstruction algorithm incorporating three-dimensional distance-dependent resolution compensation for brain SPECT: A simulation study. *Ann Nucl Med* 2002; 16: 11–18.