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Geometric calibration for a SPECT system dedicated to breast imaging*

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Abstract: Geometric calibration is critical to the accurate SPECT reconstruction. In this paper, a geometric calibration method was developed for a dedicated breast SPECT system with a tilted parallel beam (TPB) orbit. The acquisition geometry of the breast SPECT was firstly characterized. And then its projection model was established based on the acquisition geometry. Finally, the calibration results were obtained using a nonlinear optimization method that fitted the measured projections to the model. Monte Carlo data of the breast SPECT were used to verify the calibration method. Simulation results showed that the geometric parameters with reasonable accuracy could be obtained by the proposed method.

Key words: geometric calibration, acquisition geometry, tilted parallel beam (TPB), breast SPECT, Monte Carlo simulation

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1 Introduction

Dedicated breast single photon emission computed tomography (SPECT) is a nuclear imaging modality for the diagnosis of primary breast tumor [1–3]. The detector rotates around a vertical axis-of-rotation (AOR) to obtain breast images [4]. The vertical AOR orbit including the tilted head is called the tilted parallel beam (TPB) orbit. The TPB orbits allow for imaging small breast lesions and breast lesions of low activity uptake [3]

For SPECT imaging systems, the detector rotates around a static object for acquisition of projection images. Due to inevitable mechanical error and limited manufacturing accuracy, the actual values of acquisition geometric parameters cannot be identical to the designed values. However, reconstructions are very sensitive to the values of geometric parameters. Without a proper geometric calibration, the projection data could be improperly positioned leading to a loss of spatial resolution and even artifacts [5, 6]. Hence, geometric calibration is critical to obtain breast images with high image quality and improve the accuracy of breast tumor diagnosis.

It is usually difficult to directly obtain exact values of acquisition geometric parameters. In recent years, various methods using point source projections have been proposed for geometric calibration. Beque et al. proposed a method using three point sources to estimate the geometry of a pinhole SPECT with a circular orbit [7]. Wang and Tsui gave a method that determines various pinhole SPECT imaging geometries using unified projection operators in homogeneous coordinates [8]. Tran et al. adapted Beque's geometry calibration procedure to calibrate a dual-headed SPECT system with a circular orbit [9]. However, a reliable calibration method is seldom mentioned in literature for a breast SPECT with a TPB orbit. In

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this work, we developed a geometric calibration method for a dedicated breast SPECT with a TPB orbit and assessed the validity of the proposed method using Monte Carlo simulation studies.

The paper is organized as follows: Section 2 describes the proposed calibration method, focusing on the characterization of the acquisition geometry and the derivation of the projection model. Section 3 presents the simulation results. Finally, Section 4 summarizes and discusses the proposed method and future work.

2 Materials and methods

Monte Carlo simulation is widely used in nuclear medical imaging [10]. In particular, simulation provides effective way to validate new methods before performing experimental measurements. A dedicated breast SPECT was simulated. The tilted detector was mounted on the ring and rotated around the uncompressed breast with the patient in prone position. The detector consisted of a NaI crystal array and a parallel beam, lead collimator. The collimator with hexagonally arranged holes was positioned in front of the crystal array. The simulation toolkit used here was the Geant4 Application for Tomography Emission (GATE) [11, 12]. The photoelectric effect and Compton scattering were included.

2.1 Acquisition geometry

For the breast SPECT system, a tilted detector rotates on a circular orbit during acquisition. The spatial activity distribution of the object is projected through the parallel-hole collimator onto the detector. When describing the acquisition geometry, two sets of Cartesian coordinates are established. One coordinate system XYZ defines the activity distribution in the 3-D object space, with the Z axis along the AOR. The other coordinate system UV defines the projection in the 2-D detector space, with each axis along a known direction in the crystal array. The origin of the UV coordinate system is the detector center (Fig. 1).

The radius of rotation (ROR) d is the distance between the AOR and the detector center without geometric misalignment. The offset τ and η are the offsets of the UV origin from the projection of the XYZ origin, as shown in Fig. 1. The tilt Φ is defined as the angle formed by the AOR and the detector plane, and the twist Ψ is defined as the angle formed by the axis U and the axis X. In the ideal acquisition geometry, τ, η, Ψ should be zero. The set of parameters $\{d, \tau, \eta, \Phi, \Psi\}$ is necessary and sufficient to completely describe the acquisition geometry (Table 1), in which the ROR d could be determined from micrometer measurement in advance.

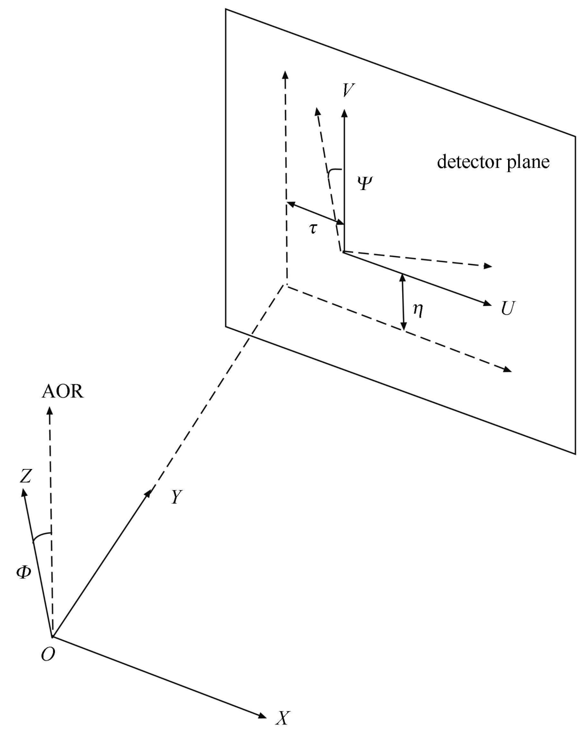

Fig. 1. Schematic of breast SPECT acquisition geometry with various parameters defined.

Table 1. Geometric parameters illustrated in Fig. 1.

symbol	description
d	radius of rotation
τ	transverse offset
η	axial offset
Φ	tilt angle
Ψ	twist angle

2.2 Projection model

This subsection gives the derivation of the projection model. Supposing a point source located in the object space is (x, y, z), the coordinates (x''', y''', z''') of the point source when the XYZ coordinate system rotated to align with the UV system is calculated as

$$\begin{pmatrix} x''' \\ y''' \\ z''' \end{pmatrix} = \begin{pmatrix} \cos\Psi & 0 & -\sin\Psi \\ 0 & 1 & 0 \\ \sin\Psi & 0 & \cos\Psi \end{pmatrix} \times \begin{pmatrix} 1 & 0 & 0 \\ 0 & \cos\Phi & -\sin\Phi \\ 0 & \sin\Phi & \cos\Phi \end{pmatrix}$$

$$\times \begin{pmatrix} \cos\theta & \sin\theta & 0 \\ -\sin\theta & \cos\theta & 0 \\ 0 & 0 & 1 \end{pmatrix} \times \begin{pmatrix} x \\ y \\ z \end{pmatrix}. \quad (1)$$

The angle θ in the above formula is the rotation angle, formed by the detector plane and the X axis. Conveniently, the Cartesian coordinates xyz are transformed into cylindrical coordinates $r\alpha z$. The above formula is rewritten as

$$\begin{pmatrix} x''' \\ y''' \\ z''' \end{pmatrix} = \begin{pmatrix} \cos\Psi & 0 & -\sin\Psi \\ 0 & 1 & 0 \\ \sin\Psi & 0 & \cos\Psi \end{pmatrix} \times \begin{pmatrix} 1 & 0 & 0 \\ 0 & \cos\Phi & -\sin\Phi \\ 0 & \sin\Phi & \cos\Phi \end{pmatrix}$$
$$\times \begin{pmatrix} \cos\theta & \sin\theta & 0 \\ -\sin\theta & \cos\theta & 0 \\ 0 & 0 & 1 \end{pmatrix} \times \begin{pmatrix} r\cos\alpha \\ r\sin\alpha \\ z \end{pmatrix}. \tag{2}$$

If the projection coordinates of the point source in the detector space is (u, v), the coordinates could be expressed in a function of the rotation angle, the acquisition geometry, and the point source location. As illustrated in Fig. 2, the projection coordinates (u, v) could be calculated as:

$$\begin{aligned} u &= x'''(\theta,\Phi,\Psi) - \tau, \\ v &= z'''(\theta,\Phi,\Psi) - \eta - d\sin\Phi. \end{aligned} \tag{3}$$

For a single-head SPECT system, there is no geometric reference for the axial XYZ origin. The axial XYZ origin could be selected to coincide with the axial UV origin and then the axial offset equals zero. Replacing η by zero yields

$$\begin{aligned} u &= x'''(\theta,\Phi,\Psi) - \tau, \\ v &= z'''(\theta,\Phi,\Psi) - d\sin\Phi. \end{aligned} \tag{4}$$

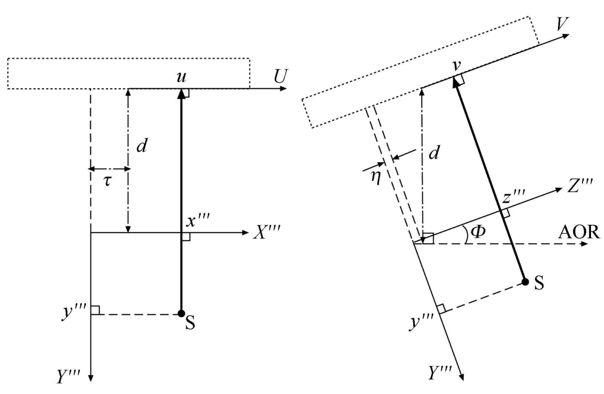

Fig. 2. Schematic of point source projection. A point source S at (x, y, z) and its projection is shown. The left shows a transverse view and the right shows an axial view.

2.3 Implementation of calibration

To determine the acquisition geometry, a calibration phantom consisting of three point sources was scanned after the actual SPECT acquisition. Projections were acquired over 360° in the step-and-shoot mode and the degree increments were identical to the priori acquisition. The projection centroid of each point source i at each angle j was calculated using the center of mass algorithm. To estimate the four unknown geometric parameters, the measured projection data were fitted to modeled projection data by minimizing a least square objective function F. The minimization of F was performed using the Powell algorithm. The initial estimates were all set to zero.

$$F = \sum_i \sum_j \Big[\big(u_{ij}^{\text{measured}} - u_{ij}^{\text{modeled}}\big)^2 + \big(v_{ij}^{\text{measured}} - v_{ij}^{\text{modeled}}\big)^2 \Big]. \tag{5}$$

In summary, the calibration was implemented in three steps: 1) derivation of the projection model, 2) acquisition of the calibration phantom projections, 3) estimation of geometrical parameters by fitting the modeled projections to the measured ones.

2.4 Simulation study

The entire system (breast SPECT and calibration phantom) was simulated using GATE. We adapted the optimal calibration phantom described in Ref. [13] to generate projection data. The projections of the calibration phantom were acquired at 4° increments over 360° in the step-and-shoot mode, as shown in Fig. 3. According to the NEMA [14] and IEC [15] standards, the proper acquisition protocol was designed to attain sufficient count statistics.

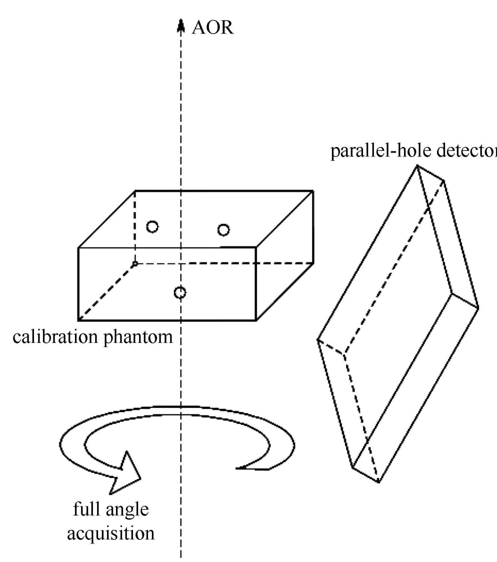

Fig. 3. Simulated setup for geometric calibration of a dedicated breast SPECT imaging system.

The goal of this study is to evaluate the performance of the proposed method. We present three simulation studies that include the tilt Φ study, the offset τ study, and the pixel size study. The entire system is modeled with three different sets of simulated parameters respectively. The designed parameters are shown in Table 2, with the parameter Ψ set to zero.

Table 2. The designed parameters' value of three simulation studies.

set	pixel size/mm	τ/mm	Φ/(°)
1	2.0	1.0	0
	2.0	1.0	30
	2.0	1.0	45
2	2.0	0.0	30
	2.0	1.0	30
	2.0	2.0	30
3	1.4	1.0	30
	2.0	1.0	30

3 Results

The centroid data and fitted curves are shown in Figs. 4–6. On the tilt Φ study (Fig. 4), the relative axial position of Point 1 and Point 2 reflects the parameter Φ. On the offset τ study (Fig. 5), the collective shift of all transverse centroids to the ideal locations is the value of the parameter τ. On the study of pixel size (Fig. 6), the centroids are closely tracked with pixels of 1.4 mm while tracking is not as accurate as with pixels of 2.0 mm.

The acquisition geometry is determined by fitting the modeled projections to the centroid data. The estimated geometrical parameters are compared with their true values. As shown in Table 3, all estimates are centered around the true values, while the spread differs for the different parameters. The observed maximum absolute error is 0.12 mm for the offset τ. For angle values, the maximum absolute errors are 0.36° for the tilt Φ and 0.49° for the twist Ψ. The results show that reasonable estimation accuracy is obtained by the proposed calibration method.

4 Discussion

In this work, we investigated the acquisition geometry of the dedicated breast SPECT with a TPB orbit. The acquisition geometry was completely characterized using the set of parameters $\{d, \tau, \eta, \Phi, \Psi\}$. Based on the characterization, the projection model and calibration process were developed.

The geometric calibration method was tested with Monte Carlo simulation data The mean values of parameter estimates show some small differences from their true values, as expected due to the parameter correlations. However, the estimation errors are below 0.5 pixel for length values and 0.5° for angle values. The errors are small compared with other contributions from the parallel-hole diameter and the detector. Therefore, reasonable accuracy has been achieved by using the proposed method.

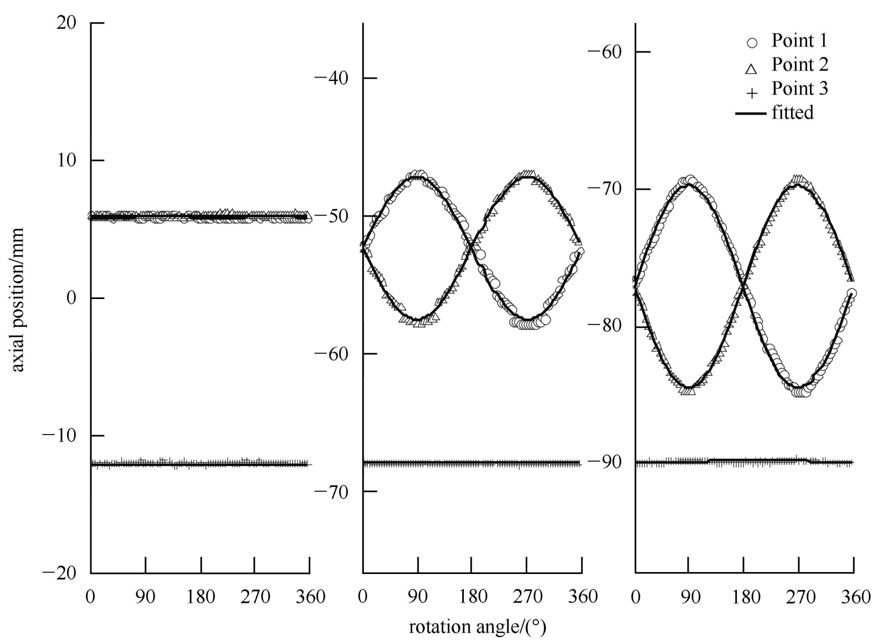

Fig. 4. Simulated values for the difference in the axial centroids for the tilt Φ study, corresponding to the parameter Φ of 0.0° (left), 30.0° (middle), 45.0° (right), respectively. The fit of (4) to data is also shown (the solid line).

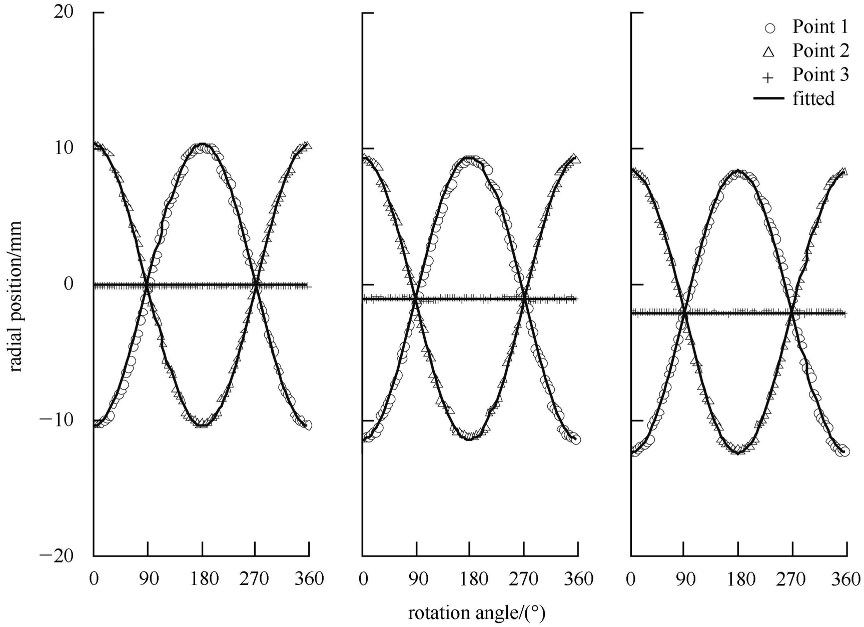

Fig. 5. Simulated values for the difference in the transverse centroids for the offset τ study, corresponding to the parameter τ of 0.0 (left), -1.0 (middle), -2.0 mm (right), respectively. The fit of (4) to data is also shown (the solid line).

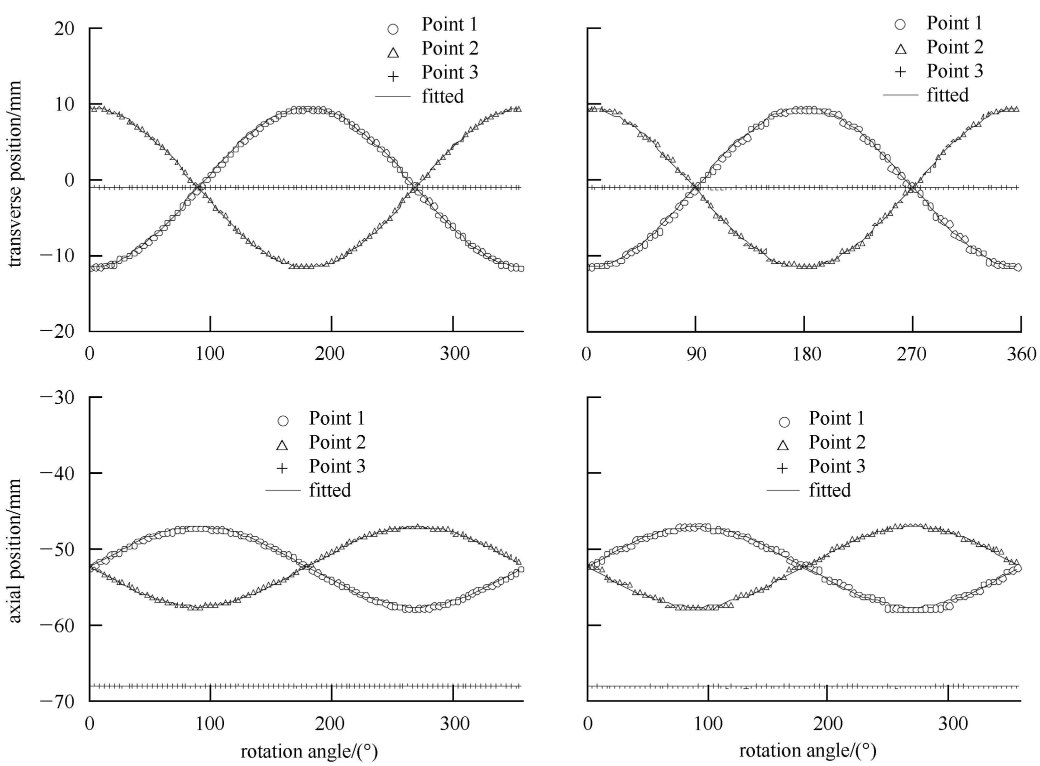

Fig. 6. Simulated values for the difference in the goodness of fit for the study of pixel size, corresponding to pixels of 1.4 mm (left column) and 2.0 mm (right column). The fit of (4) to data is also shown (the solid line).

Table 3. Calibration results for simulated breast SPECT system.

set	value	τ/mm	$\Phi/(^\circ)$	$\Psi/(^\circ)$
1	true	−1.00	0.00	0.00
	estimate	−0.90±0.01	0.24±0.09	−0.18±0.08
	true	−1.00	30.00	0.00
	estimate	−1.02±0.03	30.20±0.09	−0.18±0.04
	true	−1.00	45.00	0.00
	estimate	−0.94±0.02	45.04±0.10	−038±0.06
2	true	0.00	30.00	0.00
	estimate	−0.01±0.03	30.06±0.09	−0.10±0.06
	true	−1.00	30.00	0.00
	estimate	−1.02±0.03	30.20±0.09	−0.18±0.04
	true	−2.00	30.00	0.00
	estimate	−1.99±0.02	30.03±0.10	−0.12±0.04
3	true	−1.00	30.00	0.00
	estimate	−0.99±0.02	29.96±0.09	0.00±0.02
	true	−1.00	30.00	0.00
	estimate	−1.02±0.03	30.20±0.09	−0.18±0.04

The geometric parameters were estimated by the least square fit with the Powell algorithm. The goodness of fit was evaluated by the cost function F. Smaller F corresponds to better estimation. The study of pixel size indicates that smaller crystal size leads to a smaller F, because the accuracy of centroid calculation is improved by a small crystal size. This result shows that the proposed method provides the flexibility to calibrate higher resolution systems.

To obtain high resolution and artifact-free SPECT images, the estimated parameters with high accuracy should be used in the reconstruction process. In this paper, a geometric calibration method for dedicated breast SPECT with a TPB orbit was developed. And simulation calibrations were carried out to evaluate the projection model, the algorithm robustness, and the estimation accuracy. The calibration method has been validated by simulation studies [7]. Further, actual cameras have nonzero parameter Ψ. However, the deviation from the zero value could be expected to be small [7, 16]. In the future, the proposed method will be applied to calibrate an actual breast SPECT system. The effect of the deviation on the reconstructed images will be studied with real data.

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