Multidivergent-beam stationary cardiac SPECT

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(Received 5 January 2009; revised 23 April 2009; accepted for publication 23 April 2009; published 9 June 2009)

This article develops a stationary cardiac single photon emission computed tomography (SPECT) system using a novel multidivergent-beam collimator. This stationary SPECT system is inexpensive to build, small, and able to acquire true dynamic SPECT data. Stationary cardiac SPECT systems with multipinhole technology already exist. The proposed approach is to replace the multipinhole collimators with the originally designed multidivergent-beam collimators. The motivation for replacing the pinhole technology by divergent-beam technology is based on the following facts. The resolution/sensitivity trade-off for the pinhole is excellent (good resolution and good sensitivity) only in small object (e.g., small animal) imaging when it operates in the image magnifying mode. However, in large object (e.g., human) imaging, the resolution/sensitivity trade-off is poor (poor resolution and poor sensitivity) when the pinhole operates in the image reducing mode. In a stationary system, the number of angular views is limited; thus, image reduction is necessary to obtain more view angles. In this image reducing situation, divergent-beam collimation is able to provide better resolution and detection sensitivity than pinhole collimation. Computer simulations are carried out to verify the claims. © 2009 American Association of Physicists in Medicine. [DOI: [10.1118/1.3134245](https://doi.org/10.1118/1.3134245)]

Key words: SPECT, cardiac imaging, divergent-beam imaging, pinhole imaging

I. INTRODUCTION

Recently, manufacturers have found a large market for dedicated cardiac SPECT systems. Small field-of-view (FOV) cardiac single photon emission computed tomography (SPECT) systems have become popular due to their low cost and compact design. Spectrum Dynamics developed a D-SPECT system, which contains ten sweeping detectors. Each detector gives a set of parallel-beam measurements.¹ Digirad developed dedicated cardiac SPECT systems² that are small enough to be installed in a physician's office. These dedicated SPECT systems have relatively small gamma cameras, which are barely large enough to cover the heart. Their current detector size is 21×21 cm². In this design, the detectors remain stationary and the projections are collected as the patient, sitting upright in a chair, is rotated.

CardiArc has a system with a 180° arc of pixelated detectors with a series of lead plates (slats) to provide axial collimation for the detectors.³ A curved lead plate with a set of slits is located in front of the slats. The combination of the slits and slats provides equivalent fan-beam sampling similar to the SPRINT II.⁴ Motion of the slit plate provides the required angular sampling. The patient sits upright in a chair and is able to lean on the detector gantry as the motion of the plate is internal and is hidden from the patient. A similar slit-slat design is used in the MarC-SPECT, developed by Chang, an investigational cardiac SPECT device in which the patient rotates while the detector and the apertures remain fixed.⁵

UC San Francisco and Western Cardiology Associates developed a stationary multipinhole system for myocardial perfusion imaging.⁶⁻⁹ Since there are no moving parts, they can

perform true dynamic SPECT flow studies. In a most recent paper,⁹ the Western Cardiology Associates reported 26 patient studies using three detector, multipinhole, stationary SPECT system that had only six angular views in the transaxial direction and three views in the axial direction. In their comparison studies, they showed fewer motion artifacts in the images obtained by the stationary system than by the conventional rotational parallel-hole system.

Multipinhole collimation is the state of the art in small animal SPECT, with the main advantage being the pinhole magnification effect, which allows a high-sensitivity, high-resolution image to be obtained. Taking advantage of modern large-area gamma cameras and multidetector systems, the multipinhole technology is able to provide enough data for cardiac imaging without rotating the system gantry. A stationary system can take very fast snapshots, obtaining true dynamic imaging. The stationary system makes patient motion correction easier and is less expensive to build and maintain.

The most substantial problem faced in a stationary imaging system is the lack of sufficient view angles. In order to obtain more angular views, the pinholes, in fact, operate in an image reducing (instead of magnifying) mode. For a fixed pinhole aperture size, pinhole collimation provides excellent detection sensitivity if the object is very small and placed very close to the pinhole; however, the detection sensitivity decreases dramatically if the object is moved away from the pinhole. As the object is moved farther into the image reduction zone, where the pinhole magnification factor is less than 1, the pinhole detection sensitivity becomes worse. In this

zone, the divergent-beam collimator becomes more sensitive than the pinhole for the same specified spatial resolution.

For both pinhole and divergent-beam systems, the imaging system's FOV is determined by the detector size and the object-to-image reduction factor. If the detectors are the same and the image reduction factors are the same, both systems have the same FOV.

In a SPECT study, the organ of interest is always assumed to be in the FOV of the gamma camera; the background and other organs may be truncated, or not in the FOV, thus they are not measured. Dedicated systems are usually small, and data truncation happens frequently. Concern about data truncation has attracted the attention of researchers for many years. A recent report indicated¹⁰ that "the effect of truncation is subtle, showing a dependency upon both the distribution of activity and attenuation" and that "small inaccuracies in reconstructed images from very small FOV camera systems should have little effect on clinical interpretation." A different group reported¹¹ that "the projection truncation of the small FOV system has negligible impact on attenuation and model-based scatter correction" and "this truncation will therefore not influence the clinical diagnosis."

A potential drawback of the stationary SPECT system is the lack of a sufficient number of views. To solve this problem, the pinhole imaging system is used, operating in image reduction mode, so that many angular views of the object can be obtained at a single detector position. In order for all pinholes to see the heart, the patient must be positioned away from the collimator, although this setup reduces the resolution and detection sensitivity. The UC San Francisco group considered the trade-offs, adopted this strategy, and got encouraging observations.⁶ "Positioning the (multipinhole) collimator left anterior oblique (LAO) + right anterior oblique delivered image quality almost comparable to the parallel-hole collimator. Spatial resolution of the two collimators was comparable with the multipinhole collimator exhibiting up to fivefold higher detection efficiency at 10 cm distance."

Furthermore, they acquired SPECT data only at two detector positions, and nine pinholes were used for each position, which resulted in only six independent view angles. In Ref. 8, a Picker Prism 3000XP three-detector SPECT system was modified. The two outer detectors were positioned at a 67.5° viewing angle relative to the center LAO detector. Each detector had a six-pinhole collimator. This multipinhole system had no moving parts. They showed comparable resolution (10 mm FWHM) to a rotational SPECT system using the same detectors with low energy high resolution (LEHR) collimation. The pinhole system's detection sensitivity was four times greater than that of the comparable rotational SPECT system. That is, stationary multipinhole cardiac imaging can have comparable image resolution to parallel-hole collimators but with higher detection efficiency.

The main difference between the multipinhole system and the proposed multidivergent-beam system is the replacement of the multipinhole collimators with multidivergent-beam collimators. The advantages and disadvantages of the multidivergent-beam system compared to the multipinhole system will be discussed in the following sections.

II. THEORY

II.A. Basic expressions

Figure 1 illustrates a pinhole collimator and a divergent-beam collimator. Here we assume that the pinhole system has a focal length f_{ph} and distance b_{ph} from the focal point to the point of interest (POI). Similarly, the divergent-beam system has a focal length f_{div} and distance b_{div} from the focal point to the POI.

For a fair comparison, we will require that these two systems have the same image reduction factor

$$f_{ph}/b_{ph} = f_{div}/b_{div}, \quad (1)$$

and that the object is the same distance

$$B_{ph} = B_{div} \quad (\text{that is, } b_{ph} = b_{div} - f_{div} - L) \quad (2)$$

away from the collimator.

In order to compare these two systems, we place a small object at the POI and require that the systems give identical spatial resolutions on the detectors. Since we fix the resolution of the two systems, the superior system will provide greater geometric detection efficiency.

We then require that these two systems give identical detection sensitivities on the detectors. Since we fix the sensitivity of the two systems, the superior system will provide better resolution.

Larger detection sensitivity means that more gamma photons can be detected, and this results in lower Poisson noise in the data. Better resolution means that smaller objects (e.g., lesions) can be resolved. We use the equations from Refs. 13–15 to derive the results. We further assume that the POI is on the central axis of the pinhole.

For the pinhole geometry, we have these two relations,

$$\text{Resolution: } R_{ph} \approx d_{ph} \frac{f_{ph} + b_{ph}}{f_{ph}}, \quad (3)$$

$$\text{Geometric efficiency: } g_{ph} \approx \frac{d_{ph}^2}{16b_{ph}^2}. \quad (4)$$

For the divergent-beam geometry, we have

$$\text{Resolution: } R_{div} \approx d_{div} \frac{b_{div} - f_{div}}{L} \left(1 + \frac{1}{2} \frac{L}{f_{div}} \right), \quad (5)$$

$$\text{Geometric efficiency: } g_{div} \approx K^2 \left(\frac{d_{div}}{L} \right)^2 \left(\frac{f_{div} + L}{b_{div}} \right)^2, \quad (6)$$

where the septal thickness t is ignored for a moment, otherwise there is a $(d_{div}/d_{div} + t)^2$ factor in g_{div} .

II.B. Sensitivity comparison with the same resolution

The requirement for the two systems having the identical spatial resolution on the detectors implies

$$R_{ph} = R_{div}, \quad (7)$$

and from Eqs. (3), (5), and (7), we have

FIG. 1. Parameters in pinhole and divergent-beam systems.

$$d_{ph} \frac{f_{ph} + b_{ph}}{f_{ph}} = d_{div} \frac{b_{div} - f_{div}}{L} \left(1 + \frac{L}{2f_{div}} \right). \quad (8)$$

In order to satisfy Eqs. (8) and (1), the hole length L of the divergent-beam collimator must satisfy

$$L = f_{div} \left/ \left(\beta \cdot \frac{b_{div} + f_{div}}{b_{div} - f_{div}} - \frac{1}{2} \right) \right. \quad (9)$$

[we denote this L as L_R],

where $\beta = d_{ph}/d_{div}$.

After the resolution is specified, we can use Eqs. (4) and (6) to compare their detection sensitivities as

$$\frac{g_{div}}{g_{ph}} = \frac{K^2 (d_{div}/L)^2 ((f_{div} + L)/b_{div})^2}{d_{ph}^2 / 16b_{ph}^2}, \quad (10)$$

where K is a constant that depends on the hole shape (~ 0.24 for round holes and ~ 0.26 for hexagonal holes). If we assume that $K=0.25$, then

$$\begin{aligned} \frac{g_{div}}{g_{ph}} &= \left(\frac{d_{div}}{L} \right)^2 ((f_{div} + L)/b_{div})^2 \left/ d_{ph}^2 / b_{ph}^2 \right. \\ &= \left[\frac{1}{\beta} \cdot \frac{b_{ph}(f_{div} + L)}{b_{div}} \right]^2. \end{aligned} \quad (11)$$

Pinhole and divergent-beam collimators with the same image reducing factor can have different performances in terms of resolution. When the divergent-beam collimator hole length L satisfies Eq. (9), both collimators give the same spatial resolution on the detectors for an object at the POI. If

$$L > L_R, \quad (12)$$

the divergent-beam collimator will provide better resolution than the pinhole. Furthermore, if

$$L < L_R, \quad (13)$$

the pinhole collimator will provide better resolution than the divergent-beam collimator.

II.C. Resolution comparison with the same sensitivity

The requirement that the two systems have identical detection sensitivities on the detectors implies

$$g_{ph} = g_{div}. \quad (14)$$

From Eqs. (11) and (2), Eq. (14) becomes

$$\frac{b_{ph}(f_{div} + L)}{\beta b_{div} L} = 1 \quad \text{or} \quad \frac{(b_{div} - f_{div} - L)(f_{div} + L)}{\beta b_{div} L} = 1. \quad (15)$$

Solving for L from Eq. (15), we have

$$L = \frac{-(2f_{div} + \beta b_{div} - b_{div}) + \sqrt{(2f_{div} + \beta b_{div} - b_{div})^2 + 4f_{div}(b_{div} - f_{div})}}{2} \quad [\text{we denote this } L \text{ as } L_S]. \quad (16)$$

After the sensitivity is specified, we can use Eqs. (1), (3), and (5) to compare the resolution as

$$\begin{aligned} \frac{R_{\text{div}}}{R_{\text{ph}}} &= \frac{f_{\text{ph}}}{\beta \cdot L} \cdot \frac{b_{\text{div}} - f_{\text{div}}}{f_{\text{ph}} + b_{\text{ph}}} \cdot \left(1 + \frac{1}{2} \frac{L}{f_{\text{div}}}\right) \\ &= \frac{1}{\beta} \cdot \frac{b_{\text{div}} - f_{\text{div}}}{b_{\text{div}} + f_{\text{div}}} \cdot \left(\frac{f_{\text{div}}}{L} + \frac{1}{2}\right). \end{aligned} \quad (17)$$

Pinhole and divergent-beam collimators with the same reduction factor can have different performances in terms of detection sensitivity. When Eq. (15) is satisfied, both collimators give the same sensitivity for an object at the POI. If

$$L < L_S, \quad (18)$$

the divergent-beam collimator will provide better sensitivity than the pinhole. Additionally, if

$$L > L_S, \quad (19)$$

the pinhole collimator will provide better sensitivity than the divergent beam.

Consequently, if L is chosen in the range of $L_R < L < L_S$, the divergent-beam system will outperform the pinhole system in both resolution and sensitivity. Section II D will prove that we always have $0 < L_R < L_S$. This implies that we can always design a divergent-beam imaging geometry to outperform the pinhole system in both resolution and sensitivity.

The above conclusion is true only when the pinhole system operates in the image reducing mode. If the pinhole system is operating in the image magnifying mode (as

widely used in small animal imaging), the counterpart of the divergent-beam system is the cone-beam system. The pinhole system can outperform the cone-beam system if the object is small enough and the object is positioned close enough to the pinhole.¹²

II.D. Proof of $0 < L_R < L_S$

For any positive values of f_{div} and b_{div} with $b_{\text{div}} > f_{\text{div}} > 0$, and for practical values of $\beta > 2$, we have

$$b_{\text{div}} + f_{\text{div}} > \frac{b_{\text{div}}}{2\beta} > \frac{b_{\text{div}} - f_{\text{div}}}{2\beta},$$

$$\text{i.e., } \beta \cdot \frac{b_{\text{div}} + f_{\text{div}}}{b_{\text{div}} - f_{\text{div}}} - \frac{1}{2} > 0. \quad (20)$$

From Eq. (9)

$$L_R = f_{\text{div}} \left/ \left(\beta \cdot \frac{b_{\text{div}} + f_{\text{div}}}{b_{\text{div}} - f_{\text{div}}} - \frac{1}{2} \right) \right. > 0. \quad (21)$$

Here the assumption that $\beta = d_{\text{ph}}/d_{\text{div}} > 2$ is true because the typical value of d_{ph} is approximately 6 mm (Ref. 9) and the typical d_{div} for an LEHR collimator is about 1.1 mm and for a low energy high sensitivity collimator about 2.54 mm. This gives typical β values of 2.36–5.45, which is greater than 2.

Now we will show $L_S > L_R$, that is,

$$\frac{-(2f_{\text{div}} + \beta b_{\text{div}} - b_{\text{div}}) + \sqrt{(2f_{\text{div}} + \beta b_{\text{div}} - b_{\text{div}})^2 + 4f_{\text{div}}(b_{\text{div}} - f_{\text{div}})}}{2} > \frac{f_{\text{div}}}{\beta(b_{\text{div}} + f_{\text{div}})/(b_{\text{div}} - f_{\text{div}}) - \frac{1}{2}} \quad (22)$$

or, equivalently,

$$\begin{aligned} &\left(\frac{2f_{\text{div}}}{\beta(b_{\text{div}} + f_{\text{div}})/(b_{\text{div}} - f_{\text{div}}) - \frac{1}{2}} \right)^2 + 2(2f_{\text{div}} + \beta b_{\text{div}} - b_{\text{div}}) \\ &\quad \times \left(\frac{2f_{\text{div}}}{\beta(b_{\text{div}} + f_{\text{div}})/(b_{\text{div}} - f_{\text{div}}) - \frac{1}{2}} \right) - 4f_{\text{div}}(b_{\text{div}} - f_{\text{div}}) \\ &< 0, \end{aligned} \quad (23)$$

which can be simplified as

$$(-4\beta^2 + 6\beta)f_{\text{div}}b_{\text{div}} + (-4\beta^2 + 4\beta - 1)f_{\text{div}}^2 - (2\beta - 1)b_{\text{div}}^2 < 0. \quad (24)$$

Since $f_{\text{div}} < b_{\text{div}}$, if $\beta > 1.5$ the left hand side of Eq. (24) is upper bounded by

$$\begin{aligned} &(-4\beta^2 + 6\beta)f_{\text{div}}^2 + (-4\beta^2 + 4\beta - 1)f_{\text{div}}^2 - (2\beta - 1)f_{\text{div}}^2 \\ &= -4f_{\text{div}}\beta(\beta - 2). \end{aligned} \quad (25)$$

When $\beta > 2$, the expression in Eq. (25) is negative. In other words, when $\beta > 2$, we have $L_S > L_R$. Recall that a practical value of β is in the range of 2.36–5.45. Therefore, we always have $L_S > L_R$.

The relationship $0 < L_R < L_S$ guarantees the existence of divergent-beam collimators that are superior to the image reducing mode pinhole collimator in terms of both resolution and detection sensitivity.

II.E. A more realistic comparison between the divergent-beam and pinhole collimators

Hereafter, we address more realistic collimation situations where we consider collimator penetration and a distributed source, which may not be exactly at the center of the field of view. We assume that the radiation source is a three-

FIG. 2. Additional view angles are created by multipinhole and multidivergent-beam systems. (a) The multipinhole system creates an additional view angle θ_{ph} . (b) The multidivergent-beam system creates an additional view-angle θ_{div} . (c) If a multidivergent-beam system is positioned in 3 non-overlapping locations, it can cover over 180° of view-angles.

dimensional cube with size of $15 \times 15 \times 15 \text{ cm}^3$ containing the heart, the collimator is made of lead, and there is an angle θ between a general emission ray and the central line of the collimator. Based on these generalizations, Eqs. (3)–(6) are revised as Eqs. (26)–(29),

$$\text{Pinhole collimator resolution: } R_{ph} \approx \hat{d}_{ph} \frac{f_{ph} + b_{ph}}{f_{ph}}, \quad (26)$$

$$\text{Geometric efficiency: } g_{ph} \approx \frac{\hat{d}_{ph}^2 \cos^3 \theta}{16b_{ph}^2}, \quad (27)$$

where $\hat{d}_{ph} = \sqrt{d_{ph}[d_{ph} + 2\mu^{-1} \tan(\alpha/2)]}$ is Anger's effective pinhole diameter,¹⁶ μ is the linear attenuation coefficient of the collimator material, and α is the pinhole acceptance angle. More accurate effective pinhole diameters should consider photon penetration and have been derived in Refs. 17 and 18. For large pinholes (with a diameter larger than 1 mm), Anger's effective pinhole diameter is acceptable,^{17,18} and will be adopted in this article for its simplicity. Similarly, for the divergent-beam geometry, we have

$$\text{Divergent-beam collimator resolution: } R_{div} \approx d_{div} \frac{b_{div} - f_{div}}{\hat{L}} \cdot \frac{1}{\cos \theta} \left(1 + \frac{1}{2} \frac{\hat{L}}{f_{div}} \right), \quad (28)$$

$$\text{Divergent-beam collimator geometric efficiency: } g_{div} \approx K^2 \left(\frac{d_{div}}{\hat{L}} \right)^2 \left(\frac{d_{div}}{d_{div} + t} \right)^2 \left(\frac{f_{div} + L}{b_{div}} \right)^2, \quad (29)$$

where t is the septal thickness and $\hat{L} = (L - 2\mu^{-1}) / \cos \theta$ is the effective hole length and $K = 0.26$ for a hexagonal collimator hole.

II.F. Angular sampling advantage of the multidivergent-beam method over multipinhole

For a stationary system, the most critical concern is the lack of sufficient view angles. Both the multipinhole and multidivergent-beam systems can provide additional view angles at a fixed detector position. The additional view

angles provided by these two types of systems are different. We will use the one-dimensional (1D) version to illustrate the basic principle.

The multipinhole geometry is shown in Fig. 2(a), where the angle θ_{ph} is the additional view angle a multipinhole system provides. The maximum value of the additional view angle $\theta_{ph}^{\max}$ can be determined by

$$\theta_{ph}^{\max} = \tan^{-1} \frac{D/2 - \rho f_{ph} B_{ph}}{f_{ph} + B_{ph}}, \quad (30)$$

where D is the detector size and ρ is the radius of the object of interest.

The multidivergent-beam geometry is shown in Fig. 2(b), where the angle θ_{div} is the additional view angle provided by the multidivergent-beam system. The maximum value of the additional view angle $\theta_{div}^{\max}$ can be determined by

$$\theta_{div}^{\max} = \tan^{-1} \frac{D/2 - \rho(b_{div} - B_{div} - L)/b_{div}}{B_{div} + L}. \quad (31)$$

As a special case, in our economical cone-beam to multidivergent-beam (C2MD) collimator design, which will be presented in Sec. IV, $b_{div} = 2B_{div}$, then Eq. (31) becomes

$$\theta_{div}^{\max} = \tan^{-1} \frac{D - \rho + \rho L/B_{div}}{2(B_{div} + L)}. \quad (32)$$

A numerical example in Sec. III shows that $\theta_{ph}^{\max} < \theta_{div}^{\max}$, and each detector position using the multidivergent-beam collimator can cover approximately 60° of view angles. Therefore, three detector positions can acquire projections over 180° as shown in Fig. 2(c). On the other hand, for the multipinhole system, three detector positions are less likely to cover 180° .

This analysis shows a distinct advantage of the multidivergent-beam collimator over the multipinhole collimator: The multidivergent-beam collimator can provide a larger view-angle range than the multipinhole collimator. This analysis can be readily extended to practical two-dimensional (2D) multipinhole and multidivergent-beam collimators. The view-angle range in the axial direction is also larger for the multidivergent-beam collimator than for the pinhole collimator.

FIG. 3. A divergent-beam collimator is compared to a pinhole collimator in terms of (a) resolution and (b) sensitivity for an image reduction factor = 1/2 and distance to the collimator = 40 cm.

III. NUMERICAL EXAMPLES

III.A. Sensitivity comparison with the same resolution

We assume that the image reducing factor $f_{\text{div}}/b_{\text{div}}$ in Eq. (1) is 0.5, $f_{\text{div}}=40$ cm, and $\beta=4$, the value of L_R from Eq. (9) is $L_R=2f_{\text{div}}/(6\beta-1)=3.48$ cm. At $L=L_R=3.48$ cm, the sensitivity gain given by Eq. (11) is $g_{\text{div}}/g_{\text{ph}}=2.03$. This implies that when the pinhole and the divergent-beam systems have the same spatial resolution at the center of the object, the divergent-beam system has a twofold sensitivity gain over the pinhole system.

III.B. Resolution comparison with the same sensitivity

If we use the same system setup as in Sec. III A and assume that the image systems satisfy assumption (2), the value of L_S can be directly solved from Eq. (16) as $L_S=4.92$ cm. That is, when the collimator hole length is chosen as $L_S=4.92$ cm, both systems have the same sensitivity at the center of the object, while the divergent-beam system has better resolution than the pinhole system, with a resolution ratio of $R_{\text{div}}/R_{\text{ph}}=0.72$. In this numerical example, $L_S=4.92$ cm is rather long from a practical point of view. Thus, for a practical hole length L , the divergent-beam collimator will have better sensitivity than the pinhole.

III.C. A more realistic comparison between the divergent-beam and pinhole collimators

We now compare the divergent-beam collimator and the pinhole collimator based on the two assumptions expressed in Eqs. (1) and (2), and that the two collimators have the same reduction factor (0.5) and the same distance (40 cm) from the center of the object to the collimator. These two requirements result in $f_{\text{div}}=40$ cm, $b_{\text{div}}=80$ cm, $f_{\text{ph}}=20$ cm, and $b_{\text{ph}}=40$ cm. The pinhole diameter is $d_{\text{ph}}=6$ mm, $\alpha=90^\circ$, the divergent collimator hexagonal hole size is $d_{\text{div}}=1.5$ mm, and the septal thickness is 0.23 mm. If we assume the hole size of 1.5 mm, hole length of 3.6 cm, the linear attenuation coefficient of lead at 140 keV of 21.66/cm, then the penetration percentage is less than 6%. Using Eqs. (26)–(29), we obtained a divergent-beam to pinhole resolution ratio plot and a sensitivity ratio plot as in Fig. 3. From the curves, we have the equal resolution parameter $L_R=3.3$ cm and at this hole length the divergent to pinhole sensitivity gain is 2. From the sensitivity ratio plot, the equal sensitivity hole length $L_S=4.7$ cm and at this hole length the divergent to pinhole resolution FWHM reduction factor is 0.7.

III.D. Angular sampling advantage of the multidivergent-beam method over multipinhole

For a multipinhole system setup: $D=53$ cm, $\rho=10$ cm, $B_{\text{ph}}=40$ cm, and $f_{\text{ph}}=20$ cm, the maximum value of the additional view angle is $\theta_{\text{ph}}^{\text{max}}=19.7^\circ$ according to Eq. (30).

For an equivalent multidivergent-beam system setup: $D=53$ cm, $\rho=10$ cm, $L=3.48$ cm, and $B_{\text{div}}=B_{\text{ph}}=40$ cm, the maximum value of the additional view angle is $\theta_{\text{div}}^{\text{max}}=26.8^\circ$ according to Eq. (32). Thus at a fixed detector position, the multidivergent-beam system can provide a larger angular range than the multipinhole system.

III.E. Comparison of multidivergent-beam and multipinhole collimators via computer simulations

Two comparison studies are presented here to compare the multidivergent-beam and multipinhole imaging systems via computer simulations. In both systems, the collimators

FIG. 4. Computer simulations setup: (a) The multidivergent-beam system and (b) the multipinhole system. Both systems have the same heart-to-collimator distance B and same image reduction factor (0.5).

FIG. 5. Computer simulations results. The first row compares the angular sampling effects of (a) the multidivergent-beam system and (b) the multipinhole system. The second row compares the noise effects in (c) the multidivergent-beam system and (d) the multipinhole system. Note: In (a) and (b), the grayscale display window is shifted down on purpose in order to show the background artifacts.

had the same 2-3-2 partitions, as shown in Fig. 4 and also in Fig. 7(d). Each detector position provided five view angles in the transaxial direction. Each subdetector zone was a 64×64 matrix with a pixel size of 1.25 mm. Three detector positions were used. Both collimators had the same image reduction factor of 0.5. The adjacent detectors were positioned 60° apart. The cardiac phantom had an outside radius of 6 cm and an inner radius of 5 cm. The heart-to-collimator distance was 40 cm. In projection data generation, we assumed that these two systems had the same spatial resolution at the center of the object, which led to a twofold sensitivity gain for the multidivergent-beam system over the multipinhole system. The iterative ML-EM algorithm was used to reconstruct the images with five iterations. No resolution compensation was used in the image reconstruction.

In the first comparison study, computer simulated noiseless projections were used. The data were attenuationless and scatter-free. The purpose of this study was to compare the angular sampling effects for both imaging geometries. In both geometries, the detector partitions were the same; however, their view angles for each subdetection region were different. It is clearly shown in Fig. 5(a) that three detector positions with a multidivergent-beam collimator provided satisfactory angular sampling; the short axis reconstructions appear as circular rings. On the other hand, the same three detector positions with a multipinhole collimator did not provide sufficient angular sampling; the circular rings became a little hexagonlike and the background has artifacts in the shape of a star (see Fig. 5(b)).

In the second comparison study [see Figs. 5(c) and 5(d)], computer simulated noisy projections were used. When the

Poisson noise was added to the projections, the sensitivity gain of 2 of the divergent-beam system over the pinhole system was incorporated. The purpose of this second study was to compare the noise effects for both imaging geometries. A uniform spherical phantom with radius of 6 cm was used so that it was easier to calculate the noise standard deviation over the center region of the object. It was assumed that both systems had the same scanning time (of approximately 7 min with the patient cardiac Tc-99m dose). The multidivergent-beam system had a total photon count of 339 439, and the multipinhole system had a total photon count of 155 480. An inscribed cube inside the sphere was used to evaluate the mean and standard deviation of the reconstructed image. The normalized standard deviation (i.e., standard deviation divided by the mean) was 0.12 for the divergent-beam system and was 0.16 for the pinhole system.

IV. DISCUSSION

IV.A. C2MD conversion

There are many approaches to designing a multidivergent-beam collimator. One obvious approach is to design each divergent zone independently, which usually results in a very expensive fabrication cost. We now propose a novel and economical approach based on a cone-beam collimator.

In order to illustrate the idea, we first use a 1D example, where the cone-beam collimator degenerates into a fan-beam collimator, as shown in Fig. 6. First, we turn the collimator upside down, and the convergent-beam collimator becomes a divergent-beam collimator. Second, we partition the collimator into multiple sections (or zones), and label them as a, b, and c. Third, we cut the sections. Fourth, we rearrange and attach them in a reversed order: c, b, and a. This procedure is illustrated in Figs. 6(a)–6(e).

If the original convergent-beam collimator has a focal length f , then such a converted multidivergent-beam collimator will have a common field of view that has a distance f away from the center of the collimator. In other words, if we need to design a multidivergent-beam collimator with the center of the ROI at a distance B from the collimator, first we need to fabricate a convergent-beam collimator that has a focal length B , then we cut, rearrange, and glue to construct a multidivergent-beam collimator.

The fabrication of a practical 2D collimator can follow the same procedure, as illustrated above. That is, we start with a regular cone-beam collimator of focal length, B , then we partition and cut the collimator into sections, finally we rearrange the sections in the reversed order and glue them together as illustrated in Fig. 7.

IV.B. Zone partition on the collimator

The partition of the collimator depends on the detector size and the trade-off between detection resolution and angular sampling. For a given detector size, the use of more view angles correlates with more partitioned zones, which results in smaller projection images. The system resolution in SPECT is dominated by the collimator and the distance be-

FIG. 6. The procedure to convert a fan-beam collimator into a multidivergent-beam collimator. (a) A fan-beam collimator. (b) The upside-down collimator. (c) The divergent-beam collimator is partitioned into three sections. (d) The three sections are separated. (e) The three sections are rearranged in the reversed order and glued together to form a multidivergent-beam collimator.

tween the object and the collimator. Due to poor sensitivity of SPECT, the image on the detector cannot be too small. Our Siemens SPECT scanner detectors are 53 cm in the transaxial direction. Considering the dead area around the partitioned collimator zones, it is practical to have three zones in both the transaxial and the axial directions, resulting in a partition similar to that of the 3×3 multipinhole partition.

The 3×3 partition has a drawback in that each detector position has only three view angles. We propose to use the 2-3-2 partition shown in Fig. 7(d). We now consider the additional view angles provided by this 2-3-2 collimator partition in Fig. 8. The middle row of divergent-beam collimators has three zones, providing view angles of θ_1 , 0, and $-\theta_1$ relative to the collimator's normal direction as in Fig. 8(a),

where θ_1 is calculated as $\theta_1 = \tan^{-1}(D/(3B))$. The top row has two zones and provides view angles of θ_2 and $-\theta_1$, where θ_2 is given as $\theta_2 = \tan^{-1}(D/(6B))$, as shown in Fig. 8(b). Similarly, the bottom row provides view angles of θ_1 and $-\theta_2$. The combination of this 2-3-2 partition gives view angles θ_1 , θ_2 , 0, $-\theta_2$, and $-\theta_1$, as shown in Fig. 8(c). At $\pm\theta_1$, the data are measured twice, but at different axial view angles.

If $D=53$ cm and $B=40$ cm, then $\theta_1=24^\circ$ and $\theta_2=12.5^\circ$. Thus θ_1 is almost two times θ_2 . The view angles are almost uniformly sampled. If we use three detector positions and the adjacent detectors have an angle of 60.5° between them, then an angular range over 181.5° is almost uniformly covered. For a shorter distance B , the angular coverage is larger. For example, let $D=53$ cm and $B=30$ cm, then $\theta_1=30.5^\circ$, $\theta_2=16.4^\circ$, and the angular coverage with three detector positions will be 232.2° . If $D=53$ cm and $B=25$ cm, then $\theta_1=35.2^\circ$, $\theta_2=19.5^\circ$, and the angular coverage with three detector positions will be 269.7° . If two detector positions are used, the angular coverage is 179.8° . That is, if the distance B can be shortened to 25 cm, it may be possible to use two detector positions for cardiac SPECT imaging with the multidivergent-beam collimator.

FIG. 7. The procedure to convert a cone-beam collimator into a multidivergent-beam collimator. (a) A cone-beam collimator. (b) The upside-down collimator. (c) The divergent-beam collimator is partitioned into seven sections. (d) The seven sections are separated, rearranged in the reversed order, and glued together to form a multidivergent-beam collimator.

IV.C. Dead area around the divergent zone

We caution that the suggested multidivergent-beam collimator has a disadvantage that there are dead areas around the partitioned zones, as shown in Fig. 9(a). In Sec. II F we learn that a short B helps to increase the angular coverage. However, a shorter B results in a larger slant angle of the collimator holes causing larger dead areas in the detector. A larger B allows a larger usable detection area but a smaller angular coverage. A larger B also causes poorer spatial resolution. These trade-offs will be taken into consideration in determining the distance B of our multidivergent-beam imaging system.

A multipinhole imaging system usually suffers from a contrary problem, that is, the multiplexing problem, in which

FIG. 8. (a) The middle row of subcollimators provides relative view angles of θ_1 , 0, and $-\theta_1$. (b) The top row of subcollimators provides relative view angles of θ_2 and $-\theta_1$. (c) All three rows provide the relative view angles of θ_1 , θ_2 , 0, $-\theta_2$, and $-\theta_1$. (d) Using three detector positions, a 180° view angle can be covered.

emission rays through different pinholes may be superpositioned in the areas of the detector, as shown in Fig. 9(b). In cardiac multipinhole SPECT, the multiplexing problem always happens because the relatively small heart is in a relatively large torso. The multiplexing problem greatly reduces the information content of the projection data, which may require more iterations to reach an acceptable reconstruction or cause blurred edges of the object of interest. On the other hand, the multidivergent-beam systems never have the multiplexing problem.

FIG. 9. (a) Representation of the dead areas in a multidivergent-beam collimator. (b) The multiplexing problem associated with a multipinhole collimator.

V. CONCLUSIONS

This article presents a method to improve the current multipinhole stationary cardiac SPECT system by replacing the multipinhole collimators by multidivergent-beam collimators. We have demonstrated that in image reduction mode, the divergent-beam collimator is superior to the pinhole collimator in terms of sensitivity and resolution. The multidivergent-beam system can provide better angular sampling than multipinhole. The multidivergent-beam system does not suffer from the multiplexing problem that a multipinhole system usually has; however, the multidivergent-beam system has its dead-zone problem, which becomes severe as the focal length gets shorter. The multidivergent-beam collimator may have a much higher fabrication cost than that of a multipinhole collimator. This article suggests an economical approach to fabricate a multidivergent-beam collimator by first building a cone-beam collimator then cutting and regluing the subregions together in the reversed order. Currently a prototype multidivergent-beam collimator is under construction. When this collimator is available, phantom experiments will be carried out for comparison evaluations.

ACKNOWLEDGMENTS

This work is partially supported by the Ben B. and Iris M. Margolis Foundation and the Benning Society.

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