REFERENCES


Application of Convergent-Beam Collimation and Simultaneous Transmission Emission Tomography to Cardiac Single-Photon Emission Computed Tomography

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Single-photon emission computed tomography (SPECT) is the most commonly performed imaging technique for perfusion studies of the heart and brain. However, these organs are much smaller than the crystal surface of gamma cameras. SPECT sensitivity and resolution can be improved by using fan- and cone-beam collimators to magnify the image of these organs over a larger portion of the crystal face. Special orbits and reconstruction algorithms must be used with convergent-beam acquisitions to prevent image distortion. Differential attenuation of source activity in the chest is one of the most vexing problems in cardiac SPECT, especially with Thallium-201. Multi-headed cameras equipped with convergent-beam collimators allow a transmission image to be obtained at the same time as emission images. Applying a transmission map of the chest attenuation values to the emission images produces a true picture of source distribution in the heart. This article reviews the technical problems associated with convergent-beam geometry and simultaneous transmission emission tomography SPECT imaging of the heart.

THE DEVELOPMENT of single-photon emission computed tomography (SPECT) has had a profound influence on the practice of nuclear medicine, especially cardiac and brain perfusion imaging. The most commonly used SPECT imaging system is a rotating scintillation camera with parallel hole collimators; this system can perform both body and brain tomography as well as conventional planar imaging. However, these systems are less than ideal for perfusion imaging of the heart and brain because of the small size of these organs. Sensitivity and resolution of small organs can be improved by using a larger portion of the crystal face of large-field-of-view (LFOV) cameras. This can be accomplished with convergent-beam geometry, which magnifies the image onto the crystal surface. The most commonly used collimators for convergent-beam SPECT are fan- and cone-beam collimators, although pin- and astigmatic geometries can also be applied. The recent development of multi-headed SPECT systems offers an additional increase in sensitivity by using converging collimation on two or three detectors.

Converging collimators on multiheaded cameras also allow accurate correction of variable attenuation of source activity in the chest, a problem that plagues cardiac SPECT, especially when low energy photons of thallium-201 (201-Tl) are imaged. The increased sensitivity of these systems allows transmission computed tomography images to be obtained in a clinically reasonable time with relatively low activity transmission sources. By applying the resulting map of chest attenuation values to the emission image data, source attenuation can be corrected. The transmission images can be obtained at the same time as the emission images, a technique known by the acronym STEP (simultaneous transmission emission protocol).

CONVERGENT-BEAM COLLIMATION

In 1979 Joaszczuk et al.9 applied convergent-beam geometry to SPECT by developing a multichannel, short-bore fan-beam collimator for brain imaging. Tusi et al.10 developed a special long-bore fan-beam collimator for brain imaging that cleared the patient's shoulders, allowing the face of the collimator to more closely approach the head. Later, Joaszczuk et al.9 and Floyd et al.11 applied cone-beam collimation to brain imaging. Our group12 used similar technology to obtain the first cone-beam tomography images of the heart. Since then, we and

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other groups have applied convergent-beam geometry to multielement cameras for a further improvement in sensitivity.3

Types of Convergent-Beam Collimators

Four types of collimators are routinely used with Anger cameras: pinhole, parallel, diverging, and converging.3 Pinhole collimators work similar to a "box camera." They have a small aperture at the end of a lead cone. Gamma rays passing through the aperture project an inverted image of the source object onto the detector's crystal. The image is magnified when the distance from the source object is less than the collimator cone length; objects are magnified when further away. In the most commonly used type of collimator, parallel, the holes are parallel to one another and perpendicular to the crystal face. The gamma-ray image projected onto the crystal surface is the same size as the source object. Diverging collimators have holes that diverge from the detector face, thus creating a larger field of view with increasing distance from the collimator. They project a magnified image of the source object onto the crystal. On the other hand, converging collimators have holes that converge toward a point or line in front of the camera, thus producing a field of view that narrows with increasing distance from the collimator. Objects between the collimator face and the convergence point or line are magnified on the crystal face. The degree of magnification is related to

\[
\frac{1}{t} = \frac{O}{f} + t\left(\frac{f + t - b}{f + t + b}\right)
\]

where \( t \) is the image size, \( O \) is the object size, \( f \) is the distance from the collimator face to the convergence point, \( b \) is the distance from the collimator face to the object, and \( t \) is the collimator thickness.4,5

Fan-beam, cone-beam, and astigmatic collimators are special types of converging-beam collimators. Fan-beam collimators focus a series of points onto a line parallel to the axis of rotation of a SPECT camera (Fig 1). Cone-beam collimators are similar to pinhole collimators except rather than having the focal point at the aperture, they focus to a point beyond the object being imaged (Fig 2). Astigmatic collimators converge the gamma rays onto two orthogonal lines at different focal lengths. One focal line is parallel to the axis of rotation, the other is perpendicular (Fig 3). A fan-beam collimator is equivalent to an astigmatic collimator with the perpendicular line of focus at a distance far from the face of the collimator; a cone-beam collimator is an astigmatic collimator with both lines of focus at the same focal length. Thus, the envelope of rays produced by astigmatic geometry lies between a fan- and cone-beam shape. This shape allows nonspherically shaped objects to better fit into the field of view.6,7

A comparison of parallel, cone-beam, fan-beam, and astigmatic geometries using simulations of the Defrise phantom is shown in Fig 4. The Defrise phantom consists of seven parallel elliptical discs of identical uniform intensity spaced five voxels apart; the discs are positioned perpendicularly to the axis of rotation (Defrise M, personal communication, 1989). The parallel collimator reconstruction in the sagittal plane (Fig 4E) shows the true distribution of activity in the phantom.2 The fan-beam collimator gives equivalent results to the parallel collimator (Fig 4D), because sufficient data are acquired in one circular orbit for both geometries. The cone-beam collimator gives poor results outside the central slice (Fig 4A) because of greater slice-to-slice cross talk compared with the fan-beam and parallel reconstructions. This is because of insufficient data collection with a standard circular orbit (see below). However, note that a centrally located object is well reproduced. The astigmatic collimator results vary with the focal length of \( F_2 \). An astigmatic collimator with a

![Fig 1. Fan-beam collimator geometry.](image1)

![Fig 2. Cone-beam collimator geometry.](image2)

![Fig 3. Astigmatic collimator geometry. \( F_1 \) and \( F_2 \) are the focal lines. (Reprinted with permission.)](image3)
focal line $F_2$ equal to the cone-beam focal length and focal line $F_3$ less than the cone-beam focal length yields poor results outside the central plane (Fig 4B). However, an astigmatic collimator with $F_2$ longer than the cone-beam focal length gives better results, because it more closely approximates fan-beam geometry (Fig 4C). A fan-beam collimator rotated 90° (Fig 4F) yields the poorest results.

Cone-beam collimators produce higher sensitivity than other collimators when LFOV cameras are used. An increase of efficiency of 2.5 times over conventional parallel geometry systems with the same resolution is possible. However, SPECT studies using conventional orbits keep the cone-beam focal point in a plane. This creates artifacts, because insufficient projection data are obtained to accurately reconstruct from this type of geometry. To acquire sufficient data, the scanning trajectory of the focal point must have at least one point of intersection for any plane passing through the reconstructed region of interest. Figure 5A shows that one circular orbit does not provide sufficient data, because a plane exists that passes through the object but does not intersect the orbit. However, a circle-and-line orbit (Fig 5B), a dual orthogonal orbit (Fig 5C), and a helical orbit (Fig 5D), do satisfy the data sufficiency condition, because all planes in the object intersect with the trajectory of the focal point.

How these special orbits can be implemented at minimal cost with present SPECT equipment is illustrated in Figs 6 and 7. Figure 6 shows a rectangular detector prescanning the circle-and-line orbit by first rotating the detector through a planar orbit while the gantry remains stationary and then having the gantry translate linearly while the detector remains fixed with respect to the gantry. A helical orbit is performed by translating the gantry while simultaneously rotating the detector. The dual orthogonal orbit involves a circular orbit followed by a semicircular orbit at 90° to the first acquisition; this technique is especially applicable to brain imaging (Fig 7).

Can cone-beam studies be performed on conventional SPECT equipment using a single planar orbit? Reconstructions of conventionally acquired cone-beam studies using the Feldkamp algorithm accurately reconstruct the central plane. This is shown in image Fig 44. Phantom computer simulations show that the error in the more central slices can be kept to less than 2% if a special reconstruction algorithm is used. Clinical results of cone-beam

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**Fig. 4.** Detrise phantom with sagittal reconstructions: (a) Cone-beam, (b) Antiparallel $F_1 = 112, F_3 = 152$, (c) Parallel fan-beam, (d) Rotated fan-beam. (Reprinted with permission.)

**Fig. 5.** (A) Circular orbit. (B) Circle-and-line orbit. (C) Dual orthogonal orbit. (D) Helical orbit. B, C, and D satisfy the data sufficiency condition.

**Fig. 6.** The circle-and-line orbit. First, the camera is rotated with the gantry fixed (motion 1). Then the detector is fixed relative to the gantry, and a linear translation of the gantry is performed (motion 2). (Reprinted with permission.)

**Fig. 7.** Dual orthogonal orbit for the brain. (Reprinted with permission.)
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Fig 4. Definite phantoms with sagittal reconstructions. (a) Cone-beam. (b) Antigastigmatic (F1 = 110, F2 = 102). (c) Antigastigmatic (F1 = 112, F2 = 102). (d) Fan-beam. (e) Parallel. (f) Rotated fan-beam. (Reprinted with permission.)

Fig 5. (a) Circular orbit. (b) Plane intersection ROI. (c) Circle-and-line orbit. (d) Dual orthogonal orbit. (e) Helical orbit. B, C, and D satisfy the data sufficiency condition.

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geometry studies using conventional orbits indicate that if the heart is positioned near the central plane of the cone-beam geometry, the technique can detect lesions not seen with parallel collimators. Special Algorithms for Convergent-Beam Cardiac SPECT

Short Scan Algorithm

The optimal angular range for fan- and cone-beam SPECT acquisitions of the brain is a full 360° of data. This is not true for the heart, however. Thallium-201 has 50% of its 66- to 80-keV energy and its 140-keV photopeaks attenuated by a 1-cm-thick skull. Similarly, 99m-Tc has 50% of its 140-keV photopeaks attenuated by a 1-cm-thick skull. Therefore, 50% of the 99m-Tc has a half-value thickness of 4.6 cm for soft tissue or water. Therefore, the data are included in the SPECT reconstruction, the image quality suffers. For this reason the standard cardiac SPECT acquisition for parallel collimators is 360°, the minimum angular sampling range required to accurately reproduce the source activity distribution.

Fan-beam SPECT, on the other hand, is a limited angular range to obtain the same SPECT reconstruction, the image quality suffers. For this reason, the standard cardiac SPECT acquisition for parallel collimators is 360°, the minimum angular sampling range required to accurately reproduce the source activity distribution.

Cone-beam collimators require a special filtered back projection algorithm to allow data acquisitions less than 360°. We have termed such algorithms short scan algorithms to emphasize their optimization for a limited angular range in distinction to the Feldkamp algorithm, which is optimized for reconstruction of 360° cone-beam acquisitions.3,5

Noncircular orbits are used in SPECT cone-beam studies because orbits that closely follow the body's contour, such as the peanut orbit, can improve resolution by as much as 1 to 2 mm without any loss of system sensitivity. This is because camera/collimator resolution deteriorates with increasing distance from the organ being imaged. Unfortunately, more complex reconstruction algorithms are necessary to use noncircular orbits with convergent-beam geometry (see below).

Our short scan algorithm assumes that the central slice is sampled as a fan-beam projection data set; this is less true for slices distant from the central slice. However, the heart is positioned in the center of the field of view; this is the area we are most concerned with accurately reproducing.

The central slice of a cone-beam study will have double samples of some projection rays with 180° plus the cone-beam angle acquisitions. In other words, the same ray is sampled from two different projection angles. This produces reconstruction artifacts. Although one might think the reconstruction can be improved by setting the doubly scanned data to zero, Naprstek has shown that this technique produces severe streaks in the reconstructed image. The streaks result from the sharp cutoff of the projection data; the reconstruction function. Parker reduced the sharp cutoff by applying a smoothing window function to the sinogram image. By multiplying the projections by a window function that properly weights the doubly sampled data, the reconstruction artifacts are significantly reduced. An analytical approach works with circular orbits; however, the application of analytic methods to noncircular orbits is difficult. Here, the coordinates of the doubly sampled projections can be determined numerically. Analogous to Parker's approach, we developed a window function that eliminates the sharp boundaries; this function includes an additional fan-beam weight factor. The larger the angle, the smoother the weight function will be. The selection of the angle depends on the data and the system bandwidth. In addition, some data are not used with this algorithm; they are set to zero. The requirement of an additional range of projection angles and the loss of some of the sampled data is the price paid for an orbit-independent (works with noncircular orbits), smooth-weight function that offers an analytical solution to the double sampling problem.

Our algorithm has been tested on a computer-generated, three-dimensional heart and lung phantom without attenuation correction. The results are shown in Fig. 8. The parallel reconstruction is used as a control to determine if reconstruction artifacts are created by the cone-beam collimator.
beam geometry. For transverse slice 32, the reconstructions of the parallel, cone-beam full 360° acquisition, cone-beam short (18° + the cone angle), and the cone beam noncircular orbits (our algorithm) are equivalent. The cone-beam reconstructions off the central slice are approximations because of the nature of the incomplete data sampling. However, for a short distance off the midplane, as shown in transverse slice 35 (1.8 cm above the midplane), the cone-beam full and short scans are equivalent to the parallel reconstruction. As one moves farther away from the midplane, as shown in transverse slice 23, which is 5.7 cm anterior to the central coronal slice, truncation artifacts occur in the cone-beam reconstructions at the superior and inferior boundary. In addition, the profiles for the full and short scan cone-beam reconstructions have a lower amplitude than the profiles for the parallel reconstruction because of insufficient data sampling. Differences between full and short scan cone-beam reconstructions are more apparent in background tissue where low amplitude artifacts are seen extending horizontally from the superior and inferior walls of the heart.

The error in reconstruction with this technique has been quantified using the above model. In coronal slice 23, the maximum percent error relative to a correctly reconstructed point in the central slice is 1.4% for the cone-beam full scan, 8.4% for the circular cone-beam short scan, and 9.7% for the noncircular cone-beam short scan. However, the error for all points within the heart that are outside of the midplane (transaxial slice 32) was less than 2% for all three types of cone-beam reconstructions. Thus, although the short scan algorithm creates significant artifacts within the heart, the error for the region of interest, the heart, is small.

What is the computational power of computers interfaced to commercial SPECT cameras, reconstruction time should be within acceptable limits with future systems.18

**Truncated Projection Correction Algorithm**

Keeping the organ of interest in the narrow field of view of a converging collimator, especially a cone-beam collimator, can be a problem. An advantage of astigmatic collimators is the shape of their envelope of rays, which is intermediate between that of a fan- and cone-beam collimator. Studies have shown that the astigmatic collimator can keep nonspherical organs such as the brain in the field of view better than other types of convergent-beam collimators.14,15

Keeping the heart in the field of view with SPECT cone-beam acquisitions is especially difficult. The field of view width at any location is dependent on the cone-beam angle and collimator focal length. Broader angles and longer focal lengths make patient positioning easier. For cone-beam studies the patient is empirically positioned in the camera gantry to place the heart in the center of the field of view. Even with experience, however, truncation of the heart occasionally occurs in large patients. A more common problem is truncation of the radiopharmaceutical distribution in the tissue surrounding the organ of interest.19 When a standard reconstruction filter is applied to these truncated projections, spikes are produced in the filtered projections at the truncation edge.17,18 These spikes produce ring artifacts of apparent increased activity at the periphery of the image (Fig 9). The ring artifacts can be removed by clipping the truncation spikes.39

The truncation correction algorithm, the filtered projections are multiplied by a window function that reduces the spikes (Fig 10). The resultant data are then back-projected to form the reconstruction.

Truncation artifacts are produced in transmission images as well.9,10 Correction of truncation artifacts is more important for transmission images than emission images, because the activity in the chest outside the heart is used to create an attenuation map.10 Correction of this problem is discussed in more detail in the section on **STEP** below.

**Geometric Point Response Correction Algorithm**

The human body inside a SPECT camera's gantry can be thought of as a three-dimensional object made up of a large number of points or voxels located at various heights, widths, and depths. As stated above, the resolution of a gamma camera decreases with increasing distance from the collimator face. Therefore, the resolution of the camera for each point or voxel will vary by the distance of that point or voxel from the camera's collimator.23 This spatial variation in geometric point response causes reconstruction artifacts in SPECT images such as shape distortion and nonuniform density variations.23,24

Geometric point response problems occur in parallel as well as convergent-beam collimation. In fact, Kuhl27 recognized the problem early in the history of SPECT; he attempted to reduce the effect of point response deterioration by performing half back projections, ie, back projecting only half way through the image array. This technique was later modified so that the back projected value weight was increased for pixels close to the detector and decreased for pixels more distant.24,25 Others have attempted to deconvolve the point response from projection data using variations of Wiener filtering where the modulation transfer function is approximated for a point source located at a specific distance.25 However, because the geometric point response varies spatially, the most accurate, although computationally demanding, technique is to model the spatially varying...
geometric point response and solve the reconstruction problem by determining the solution to a large number of linear equations.\textsuperscript{20} Other investigators have modeled the geometric point response variation as a two-dimensional point response function.\textsuperscript{20,42} Our group and Tsu's extended that work to three dimensions for use with parallel, fan- and cone-beam reconstructions.\textsuperscript{43}

In this technique, models for varying geometric point response, as well as photon attenuation and scatter, are incorporated into the projector-back projectors used as part of the iterative EM reconstruction algorithm.\textsuperscript{44} We have taken two different approaches. In the first, the spatially varying geometric point response is modeled as a three-dimensional inverse cone of rays that emanates from each detector bin (in other words, the cone projects from the detector face toward the radiocactivity); this in essence transforms each parallel collimation into a cone-beam reconstruction problem.\textsuperscript{24} The projection and back projection operations sum along each ray, weighting the contribution to the sum by the geometric response attenuation factors that are calculated during the projection and back projection summation along each ray. This technique is computationally very intensive; it was originally tested with 50 iterations on an IBM (Boca Raton, FL) 3090-600S supercomputer. A second, more computationally efficient approach is to use the shift invariant property of the collimator geometric response at a fixed distance from the detector.\textsuperscript{5,6} With this model the attenuation factor is calculated along the central ray of the point response for a sampled bin as a function of the attenuating material between the point of interest and the detector, and the attenuation is assumed to be constant over the spread of the point response for that bin at a fixed distance from the collimator face.

The effect of the algorithm for the Jaszczak cardiac phantom is shown in Fig. 11.\textsuperscript{20} Here the improvement that occurs for parallel, fan-beam, and cone-beam geometries is shown. Note the significant improvement in resolution with geometric response correction and attenuation compensation; geometric point response correction by itself produces a smoothing of the images without any loss in resolution. Figure 12 shows the effect of correcting for geometric point response in a cone-beam study of a human heart.

Geometric Parameter Estimation Algorithms

For all types of collimator geometries, it is important to precisely measure the geometric parameters of the imaging system so that the camera, collimator, and rotating gantry are accurately oriented in space relative to the center of rotation. In a perfect system, a point source placed at the center of a detector using a 64 × 64 matrix would be at coordinates 32.5, 32.5 at any camera rotation angle. In practice, the alignment varies with angular camera position because of a variety of factors including degradation of the rotating gantry's mechanics, changes in crystal packing with resulting shifts in crystal placement, and electronic shifts in camera tuning and ADC analog-to-digital converter gain.\textsuperscript{52} An offset or error factor can be calculated and used to correct minor discrepancies. Without correction, reconstruction artifacts can, because since the projection rays are not always placed in the correct projection bin.

Determining the center of rotation is relatively easy for parallel-collimated SPECT systems. A point or line source is placed within the central region of the rotational field of view, and an even number of paired 180° opposed projection images are obtained. The center of rotation for each complementary pair of images is obtained by

\[
\text{COR} = (P_x + P_y) / 2
\]

where \(P_x\) and \(P_y\) are the maximum pixel element in the opposing views. The values from each set of opposing projection images are then averaged.\textsuperscript{14}

With fan-beam and cone-beam geometry, two geometric parameters are important for accurate reconstructions: the center of rotation and the collimator focal length.\textsuperscript{14} The collimator focal length needs to be measured only once because it is fixed. As with parallel systems, the center of rotation must be measured periodically.

The center of rotation and necessary correction factor cannot be calculated as easily for convergent-beam geometry as it is for parallel collimators. We have developed a method for fan-beam geometry that includes measurement of focal length and several other parameters and have extended that work to cone-beam collimators.\textsuperscript{44} With both fan-beam and cone-beam geometry, the measured projections are not a linear function of the geometrical parameters. Therefore, nonlinear techniques are needed to estimate the parameter values.

For this method a point source is placed in the field of view of the camera away from the center of rotation.\textsuperscript{44} Projections of the point source are collected over 360° and are digitized into 64 × 64 or 128 × 128 matrices. The coordinates of the centroids of the projected point source are calculated over 360°; the Marquardt algorithm is then applied to fit the measured centroid coordinates to nonlinear expressions of the geometrical parameters. A newer technique using five point sources arranged in two intersecting lines better estimates the geometric parameter in that a unique solution can be determined for any initial solution given to the iterative estimation algorithm.\textsuperscript{9}

The algorithm also can be used to measure...
the accuracy and precision of the manufactured collimator during acceptance testing. We have found significant variation in the estimated collimator focal length in some collimators when point sources are placed at different positions in the field of view. These variations result from deviation in the assumed collimator hole angulation from the design values in individual collimator holes. Our studies of fan-beam collimators with long focal lengths (72 cm) have shown that focal length can vary significantly without causing reconstruction artifacts. However, as collimator focal length shortens, tolerance for variation diminishes.

Algorithm for Spatially Varying Focal Length Collimators

To overcome the truncation produced by conventional fan-beam collimators, a new type of fan-beam collimator has been proposed that has a spatially varying focal length over the collimator face.23,26 The center of the collimator has a short focal length that provides good magnification of the organ of interest. The focal length at the edge of the collimator is longer, reducing truncation of peripheral structures (Fig. 13). If the minimum focal length (a) of the collimator is less than the distance from any part of the patient to any detector location, a double image is produced on the detector for those distributions located with distances shorter than a.

Use of spatially varying focal length fan-beam collimators requires a special reconstruction algorithm. A technique developed here and at Duke University assumes that the variation of focal length increases monotonically toward the edge of the collimator.24 The algorithm is expressed as an infinite series of convolutions followed by one back projection. Computer simulations show that only a small number of finite terms (N) are needed to obtain high-quality reconstructed images. The weighting and convolution are executed N times, where N is the total number of projections. Eight observers viewed reconstructed transaxial images from parallel, cone- and fan-beam collimator geometries and indicated the degree of certainty with which they detected Gaussian-shaped defects at various locations. Receiver operator characteristic (ROC) analysis showed that the cone-beam design was superior to the fan-beam and parallel collimators regardless of slice position. The fan-beam outperformed the parallel collimator. Results for the cone-beam collimator showed a probable decrease in lesion detection in noncentral slices. Clinical images have been excellent with fan- and cone-beam collimators (Figs. 14 and 15).

SIMULTANEOUS TRANSMISSION EMISSION TOMOGRAPHY

One of the most vexing problems in cardiac SPECT imaging is attenuation of source activity. Attenuation artifacts reduce the accuracy of both visual and computer-assisted interpretation of perfusion images and prevent accurate quantification of radiopharmaceutical uptake.13,14

Two approaches have been taken to correct for attenuation in SPECT images. The first assumes uniform attenuation in the thorax.28,29

The chest, however, is composed of several tissues whose attenuation properties vary greatly. Therefore, these algorithms create incorrect attenuation coefficients, producing an inaccurate picture of cardiac radiopharmaceutical activity.4 The second approach uses transmission computed tomography (TCT) to directly measure the attenuation coefficients of the different chest tissues.38 The transmission scan is usually performed with a radionuclide line or point source, although x-ray tubes have also been used.46 The TCT and emission scans can be performed at separate times (sequential scanning) or at the same time (simultaneous scanning). Transmission scanning is technically easier to perform sequentially. When simultaneous acquisitions are performed, the emission and transmission data can be contaminated by photons from the other source. However, separate transmission and emission acquisitions usually increase both the imaging time and the risk of misregistration of the TCT and SPECT images resulting from patient misalignment or motion.3 A fast TCT technique has been developed by Tanaka et al.47 that allows a separate transmission scan to be performed in 2 to 12 minutes using a single source; theoretically, the scan time could be reduced to as little as 40 to 240 seconds if three line sources were used on a triple-headed camera.

Attempts have been made to perform transmission and emission SPECT scans simultaneously. Initial studies were done on single-headed systems using parallel collimators; a flood transmission source was mounted opposite the camera.36,48 This approach has a number of problems. Cross-talk between the transmission and emission data is significant. Parallel systems have relatively poor counting statistics that require high activity flood sources or long imaging times; both result in a higher radiation burden for the patient. Finally, the weight of the planar flood source is significant, making construction of a usable source holder difficult.

Multiheded cameras using converging collimators have a number of advantages over single-headed parallel systems for the performance of simultaneous transmission emission tomography.4 With three detectors, techniques can be used that compensate for transmission-emission data crosstalk. The increased sensitivity of fan-beam collimation over parallel geometry allows the use of transmission sources of relatively low activity without lengthening the acquisition. Finally, the transmission source is a small...
A line source is used as the transmission source with our fan-beam collimators; a point source can be used with cone-beam collimators. Our line-source is 58.8 cm in length and 1.2 mm in diameter; it is collimated to have an aperture of 40°.

The radiotrace used as the transmission source with STEP should have a photopoint energy that is different from that of the emission source. It is best if all the photopoints energies of one source are lower than the other source. With this arrangement, at least one source will not be contaminated by downscount from the other. The energy spectrum per disintegration of $^{133}I$ is 94.4% 66- to 80.3-keV mercury k-rays, 2.7% 133-keV gamma rays, and 10% 107-keV gamma rays. This, an adequate (although not ideal) low-cost transmission source is $^{133}I$ with its mononenergetic, 140-keV gamma photon energy (88% abundance).

When performing STEP using $^{133}I$ cardiac perfusion agents such as estamibi, gadolinium-DTPA with an approximate energy level of 99 keV can be used. Gadolinium is expensive and is not readily available. Technetium-99m can be used as both the emission and transmission source if one detector is used exclusively for transmission imaging; this means emission data from one detector is lost. The $^{133}I$ transmission source activity needed to perform STEP with a $^{133}I$ emission source is 10 to 20 mCi. Transmission imaging quality improves at higher activity levels, but it results in more corruption of emission data. We estimate the radiation exposure to the patient from the transmission source to be 10 mrad (0.0001 Gy) when 20 mCi of $^{133}I$ is used.

Collimators

Two sets of three low-energy, high-resolution fan-beam collimators have been used. One set has a focal length of 50 cm, the other 65 cm. The shorter the focal length, the less the transmission of the transmission data, especially in large patients (Fig. 17). However, if the focal length is too long, the incident photon beam from the transmission source can be truncated by the adjacent detector head as it moves inward during the body contour orbit (Fig. 18). In practice we have found the longer focal length to be superior.

Clinical Acquisition Protocol

Sixty-five- or focal length fan-beam collimators and a 15-mCi $^{133}I$ transmission source are used. Two pulse-height analyzer windows are set. One 15% window is centered at 140 keV for the transmission image; a second window of 30% is centered at 73 keV for the emission data.

After exercise or pharmacological stress, the patient is injected with 3 mCi of $^{133}I$; a second I-mCi injection of $^{133}I$ is done at redistribution.

Each detector acquires two sets of projection data, one for each energy window; activity is corrected for decay. Data are acquired over an angular range of 240° for each detector. There are 80 stops, with a 15-second acquisition at each stop.

Transmission Data

Corrector correction. A significant number of $^{133}I$ and 167-keV gamma photons from $^{133}I$ are detected in the 140-keV $^{133}I$ window of the camera head acquiring the transmission data. If the contamination is not corrected, the calculated attenuation coefficients for the heart will be incorrectly lowered. An estimate of $^{133}I$ and 167-keV photon contamination into the $^{133}I$ window can be made using the two detectors that are acquiring emission-only data. An average of the 140-keV data for these two detectors is taken at each projection. Using these data, the $^{133}I$ and 167-keV photon contamination is subtracted from the 140-keV window of the detector that is imaging transmission data before the attenuation coefficient distribution is reconstructed.

Transmission correction. The magnification of the fan-beam collimators causes some of the patient's chest to be missed or truncated. The transmission reconstruction problem was formulated such that the distribution of the attenuation coefficients was determined from the system of linear equations for only the measured projections. An iterative EM algorithm was used to solve this underdetermined system of linear equations by maximizing the likelihood function. The algorithm was constrained to reconstruct the attenuation distribution on a finite support. Because the detectors follow a body contour orbit that closely hugs the body, the detector
The transmission data are converted to projections of linear attenuation coefficients by taking the natural logarithm of the ratio of the incident flux to the measured transmitted counts. The resulting projections are reconstructed using the EM algorithm. The initial solution for the reconstruction is set to one inside the elliptical support and zero outside. Because the EM algorithm is multiplicative, the zero setting constrains the reconstruction to the elliptical support.

Because the reconstructed attenuation coefficients are energy dependent, the final step is to convert the attenuation coefficients calculated for the 140-keV transmission source and the 73-keV energy level of 201Tl. This is done by assuming that the attenuation coefficient varies linearly between 140 keV and 73 keV. The calculated 201Tl factors are then stored in memory for application to the emission data.

radius can be used to approximate a semielliptical body contour. An estimate of the support was obtained using detector radius information provided by the system software.

**Emission Data**

**Cosmwook Correction**

Transmission photons backscatter and forwardsscatter into the 73-keV 201Tl windows of the three detectors. The backscatter and forwardsscatter must be dealt with differently. When the line source was positioned as in Fig 18, it was found that in most projections the profile of the backscattered photons decreased rapidly in the direction away from the line source for the two emission-only detectors. The scatter profile in the data array at position one did not overlap with the profile in the data array at position two. Using this fact, a technique was developed to remove the contaminating backscattered transmission photons as illustrated in Fig 19.

First, the emission projection measurements are added together from the two emission-only detectors. Then the emission data from one of the detectors is subtracted from the emission data of the other. The uncontaminated emission data are obtained by taking the difference between the summed image and the absolute value of the difference image. Because the angular range of the scan is less than 360°, some projections do not have complementary views. For these projections the backscatter profile is estimated from other views with similar anatomic projections to those not measured. This technique corrects the backscattered contamination into the two emission-only detectors.

The forwardsscatter into the detector opposite the transmission source that is collecting both transmission and emission data is eliminated using a method developed by Frey et al. A fraction of the cross talk-corrected transmission photons in each projection bin of the 140-keV window are subtracted from the projection data for the 73-keV emission window.

\[ E_r = E_{in} - f_x \times T \]

where \( E_r \) is the projection data from a bin measured in the 73-keV emission window and \( f_x \) is a fraction of the cross talk-corrected transmission photons \( T \) in the same projection bin as \( E_r \), but in the 140-keV window. The fraction \( f_x \) is determined by the water-equivalent thickness of the absorber along the particular projection ray for that projection bin.

**Reconstruction**

The corrected emission projection data are reconstructed using 20 iterations of the EM algorithm. To speed computer processing, the attenuation factors are retrieved from RAM during the projection and back projection opera-
short-axis views with and without attenuation correction. Note the improved visibility of the lesions; however, there is increased background noise. Figure 22 shows profiles of a short-axis slice from a patient study with and without attenuation correction. An increase in signal after attenuation correction occurs.

We have compared the utility of attenuation correction using STEP in routine SPECT images in 25 patients being evaluated for coronary artery disease with $^{99m}$Tc studies. The images were reviewed by three trained observers who blindly and randomly interpreted the routine SPECT and STEP images. ROC analysis found a statistically significant advantage of STEP compared with routine SPECT for the diagnosis of coronary artery disease ($P < .001$). Transmission and emission images from a representative case are shown in Fig 23. Convergent-beam collimation significantly improves the sensitivity for cardiac SPECT. The technique requires special algorithms to create optimal reconstructed images. Fan-beam collimation combined with multileafed cameras allows simultaneous transmission emission tomography. STEP performs attenuation correction by direct measurement of the variable attenuation in the chest. The resulting corrected emission images are superior to and more quantitatively accurate than routine images for detection of coronary artery disease.

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