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Review of convergent beam tomography in single photon emission computed tomography

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Abstract. Investigation of convergent-beam single photon emission computed tomography (SPECT) is actively being pursued to evaluate its clinical potential. Fan-beam, cone-beam, pin-hole and astigmatic collimators are being used with rotating gamma cameras having large crystal areas, to increase the sensitivity for emission and transmission computed tomography of small organs such as the thyroid, brain or heart. With new multi-detector SPECT systems, convergent-beam geometry offers the ability to simultaneously obtain emission and transmission data necessary to quantify uptake of radiopharmaceutical distributions in the heart. The development of convergent-beam geometry in SPECT requires the integration of hardware and software. In considering hardware, the optimum detector system for cone-beam tomography is a system that satisfies the data sufficiency condition for which the scanning trajectory intersects any plane passing through the reconstructed region of interest. However, the major development of algorithms has been for the data insufficient case of single planar orbit acquisitions. The development of these algorithms have made possible the preliminary evaluation of this technology and the imaging of brain and heart are showing significant potential for the clinical application of cone-beam tomography. Presently, significant research activity is pursuing the development of algorithms for data acquisitions that satisfy the data sufficiency condition and that can be implemented easily and inexpensively on clinical SPECT systems.

1. Introduction

In industrial and medical applications convergent-beam geometry (figure 1) has become an important tomographic technique that is useful in obtaining solutions of inverses problems with better resolution, sensitivity and faster data acquisition times than with parallel geometry. Jaszczak et al (1979) originally applied convergent-beam geometry to increase the sensitivity for single photon emission computed tomography (SPECT) of the brain by utilizing a multi-slice, short-bore fan-beam collimator that focuses to a line oriented parallel to the axis-of-rotation of a rotating gamma camera. This was followed by a specially designed, long-bore fan-beam collimator for imaging the brain that would allow the face of the collimator to approach near the head and still allow the camera to clear the patient's shoulders (Tsui et al 1986). Later, Jaszczak and coworkers (Jaszczak et al 1986, 1988a, b, Floyd et al 1986a) implemented cone-beam collimation to further optimize the sensitivity of rotating gamma cameras for brain
The purpose of using fan- and cone-beam collimators is to utilize more of the available crystal area.

SPECT. This technology was applied by our group to obtain the first cone-beam tomography images of the heart (Gullberg et al 1991a, b). More recent developments of multi-detector SPECT systems offer an additional increase in sensitivity by combining converging collimation with multiple detectors. Also, these new multi-detector SPECT systems offer the ability to simultaneously obtain emission and transmission data necessary to correct for the variable attenuation in the chest in reconstructing the uptake of radiopharmaceutical distributions in the heart (Tung et al 1991a).

To fully develop convergent-beam tomography, as well as to identify its strengths and potential practical applications, requires research which simultaneously develops and optimizes both hardware and software. For certain applications, algorithms can produce good image quality and quantitative information in spite of hardware limitations that are imposed by the physics of the imaging process or that are due to design trade-offs influenced by cost considerations. In some cases new algorithms are needed because of scanning geometries that are used to improve the physical properties of the imaging detection process such as the use of body contouring orbits to minimize the fall-off of the geometric point response with distance from the collimator face, or 'short-scan' acquisitions to minimize attenuation artifacts. In other cases algorithms are able to minimize reconstruction artifacts for those SPECT systems which, because of cost constraints, cannot acquire sufficient data for cone-beam tomography. Depending upon the SPECT system, certain potential exists to develop algorithms that can produce good image quality and quantitative information in spite of hardware limitations and trade-off considerations.

This paper reviews recent developments in hardware and algorithms in the application of convergent-beam tomography to clinical nuclear medicine. Phantom and patient
results are presented showing the potential for the use of this technology in cardiac and brain imaging.

2. Hardware considerations

2.1. Rotating gamma camera

Today, the rotating scintillation camera is used more widely than fixed-ring systems because of its ability to perform both body (figure 2) and brain (figure 3) SPECT as well as conventional planar imaging. Most of the SPECT systems are single detector

Figure 2. Cone-beam collimation increases sensitivity for heart imaging. Single circular orbit scans do not satisfy the data sufficiency condition but the reconstructions are of good image quality because the heart can be positioned near the central plane of the cone beam geometry.

Figure 3. Tilted cone beam tomography of the brain can give high-resolution transaxial images. However, elongation near the top of the brain is seen in sagittal and coronal images because the tilted cone-beam geometry places these structures further from the central plane of the cone-beam geometry.
but recently systems with three (figure 4) and four rotating detectors have been commercially made available. Several manufacturers are also selling large field-of-view dual camera systems that offer whole-body imaging and computed tomography capability. With convergent-beam collimators, these systems offer an advantage of a significant increase in sensitivity for imaging small organs because of the large field-of-view of up to 60 cm compared with the standard 40 cm detectors.

The new three-detector SPECT systems (figure 4) offer a unique detector arrangement, that lends itself to easily acquiring simultaneous transmission and emission computed tomography studies, and make it possible to obtain a measure of the distribution of attenuation coefficients in an acceptable clinical imaging time. A transmission source with either a different energy than or the same energy as the emission source is positioned at the focal line or focal point, depending upon the collimator, opposite one of the detectors. This detector acquires both transmission and emission data while the other two detectors acquire only emission data. The projection data are pre-processed to subtract out cross-talk between the transmission and emission data sets. Fan-beam and cone-beam collimation improves the sensitivity for both emission and transmission studies and significantly reduces the source strengths required for the transmission scans. The combination of multiple detectors with converging collimators offers for the first time a reasonable approach to correcting for the variable attenuation in the thorax which has constantly plagued cardiac SPECT imaging.

2.2. Collimation

Several different convergent-beam collimator geometries have been used in SPECT with rotating gamma cameras. These include fan-beam (figure 1), cone-beam (figures 2 and 3), pin-hole (Palmer and Wollmer 1990) and astigmatic (figure 5) geometries with the envelope of rays arranged symmetrically or asymmetrically (Jaszczak et al 1986) about

![Figure 4](transmission_source.png)

Figure 4. Multi-detector SPECT systems increase the sensitivity and also offer the potential to simultaneously collect emission and transmission data. The detector opposite the line source collects both transmission and emission data, whereas the other two heads simultaneously collect only emission data.
Figure 5. The astigmatic geometry has two lines of focus. One has a focal length of $F_1$ and the other has a focal length of $F_2$.

a central axis that is oriented either orthogonal (figure 2) or oblique (figure 3 (see also Manglos 1989, Manglos et al 1989b)) to the axis of rotation. In particular, the astigmatic collimator (Hawman and Hsieh 1986) is a generalization of the fan- and cone-beam collimators with two orthogonal lines of focus at different focal lengths. One focal line is aligned parallel and the other is aligned perpendicular to the axis of rotation. A fan-beam collimator is equivalent to placing the perpendicular line of focus at a distance far from the face of the collimator, whereas a cone-beam collimator is an astigmatic collimator with both lines of focus at the same focal length. As both focal lengths increase, the astigmatic collimator approaches that of a parallel collimator.

Simulations of the Defrise phantom (Defrise 1989) in figure 6 are shown in figure 7 comparing cone-beam, fan-beam, parallel and astigmatic geometries. A generalized astigmatic reconstruction algorithm was applied to the simulated projection data generated over one complete rotation about the axis of rotation in figure 5. The astigmatic reconstruction algorithm uses the same ramp filtering in the transaxial direction as the Feldkamp cone-beam reconstruction algorithm (Feldkamp et al 1984) but backprojects along an astigmatic geometry. In each simulation this algorithm was applied by setting the focal lengths appropriate for the astigmatic, fan-beam ($F_1 = 112$, $F_2 = 9999$), cone-beam ($F_1 = 112$, $F_2 = 112$) and parallel geometries ($F_1 = 9999$, $F_2 = 9999$), where all dimensions are in units of voxels equal to 0.6 cm.

The simulations (figure 7) illustrate that the cone-beam reconstruction has more slice-to-slice crosstalk in the sagittal reconstructions of the Defrise phantom than the fan-beam and parallel reconstructions (see also Zeng and Gullberg 1990a). The parallel and fan-beam geometries give the best results and are expected to be equivalent since sufficient data are acquired in one circular planar orbit for these geometries. A comparison of the astigmatic ($F_1 = 112$, $F_2 = 72$) and cone-beam reconstructions indicates that an astigmatic geometry with the focal line $F_1$ equal to the cone-beam focal
Figure 6. Defrise phantom consists of seven parallel ellipsoidal discs of identical uniform intensity equally spaced at 5 voxels apart parallel to the axis of rotation.

Figure 7. Sagittal reconstructions of the Defrise phantom for (a) cone-beam ($F_1 = 112, F_2 = 112$), (b) astigmatic ($F_1 = 112, F_2 = 72$), (c) astigmatic ($F_1 = 112, F_2 = 152$), (d) fan-beam ($F_1 = 112, F_2 = 9999$), (e) parallel ($F_1 = 9999, F_2 = 9999$), and (f) rotated fan-beam ($F_1 = 9999, F_2 = 112$) geometries.
length and the focal line $F_2$ less than the cone-beam focal length will give poor results. However, an astigmatic geometry ($F_1 = 112, F_2 = 152$) with $F_2$ longer than the cone-beam focal length gives better results because it more closely approximates a fan-beam geometry. Also, we see that the rotated fan-beam geometry gives the worst results of all the simulations.

2.3. Orbits

Cone-beam collimation has the best sensitivity for large crystals but, as illustrated in the simulations in figure 7, does not produce artifact-free reconstructions using a scanning trajectory whose focal point remains in a plane. Presently, all SPECT systems acquire data by scanning only in single planar orbits where the focal point traverses either circular or elliptical curves lying within a single plane (figure 8(a)). These data acquisitions have been shown to give insufficient projection data (Tuy 1983, Smith 1985). The reconstruction, using the Feldkamp algorithm (Feldkamp et al 1984), is an approximation except for the central plane which is reconstructed accurately (figure 7). Preliminary results of cone-beam tomography of the heart (Gullberg et al 1991a, b) look promising when the heart is positioned near the central plane of the cone-beam geometry and demonstrate improved lesion detection. These conclusions are based on a limited number of clinical studies and subjective evaluation by clinical investigators.

For cone-beam tomography, we know that a detector system will acquire sufficient data (Tuy 1983, Smith 1985) if the scanning trajectory of the focal point has at least

![Image](https://example.com/image.png)

**Figure 8.** Data sufficiency condition. (a) One circular planar orbit does not satisfy the data sufficiency condition since a plane can be found that passes through the object but does not intersect the orbit. However, (b) a circle-and-line orbit, (c) a dual orthogonal orbit, and (d) a helical orbit do satisfy the data sufficiency condition.
Figure 9. The circle-and-line orbit illustrated in figure 8(b) can be implemented by first rotating the camera with the gantry fixed (motion 1) and then fixing the detector with respect to the gantry and performing a linear translation of the gantry (motion 2).

Figure 10. Illustration of the application of a dual orthogonal orbit in figure 8(c) to cone-beam tomography of the brain.
one point of intersection for any plane passing through the reconstructed region of interest. In figure 8(a) we see that a planar circular orbit does not satisfy this data sufficiency condition since there is a plane that passes through the object but does not intersect the orbit. However, non-planar orbits such as a circle-and-line orbit (figure 8(b)), a dual orthogonal orbit (figure 8(c)) and a helical orbit (figure 8(d)) do satisfy the data sufficiency requirement.

These orbits could be implemented to satisfy the data sufficiency condition with present SPECT systems at little additional cost. Figure 9 illustrates the implementation of the circle-and-line orbit using a rectangular detector that first rotates through a planar orbit while the gantry remains stationary and then the gantry performs one linear translation while the detector remains fixed with respect to the gantry. A helical orbit can easily be implemented by translating the gantry while simultaneously rotating the detector. The dual orthogonal orbit would be particularly applicable for brain imaging and could be implemented using one transaxial orbit and one half orbit that scans over the top of the head as illustrated in figure 10. In looking at ways to improve cone-beam tomography, we need to consider the gain obtained in image quality with sufficient angular sampling versus the cost of additional necessary hardware.

3. Algorithm considerations

The application of various converging collimator geometries in clinical SPECT has required the development of new algorithms to solve several technical problems. Here we review the development of these algorithms that have made possible the clinical implementation of convergent-beam tomography using present clinical SPECT systems.

3.1. Non-circular filtered backprojection cone-beam reconstruction algorithm

During a patient scan the camera rotates in a non-circular orbit, with the centre of the camera always pointing to the centre of rotation, and closely follows the patient’s body contour to minimize the fall-off in resolution with distance from the collimator. New fan-beam (Weinstein 1980) and cone-beam (Gullberg et al 1991b, Gullberg and Zeng 1992) reconstruction algorithms were developed to reconstruct projections from non-circular detector orbits. A cone-beam reconstruction algorithm (Gullberg et al 1991b, Gullberg and Zeng 1992) was derived using the non-circular fan-beam algorithm developed by Weinstein (1980) and extending it to cone-beam geometry using the method of Feldkamp et al (1984) to derive the cone-beam reconstruction algorithm for non-circular planar orbits.

3.2. ‘Short-scan’ filtered backprojection reconstruction algorithm

In the application of SPECT to cardiac imaging, it was found that the reconstruction of the posterior projections of emissions from a radiopharmaceutical localized in the heart resulted in reconstruction artifacts (Coleman et al 1982, Tamaki et al 1982, Go et al 1985, Eisner et al 1986, Tsui et al 1989b). Therefore, an angular sampling range is usually chosen so as not to acquire the highly attenuated posterior projections. For fan- and cone-beam geometries new ‘short-scan’ reconstruction algorithms (Zeng and
Gullberg 1990b, 1991, Gullberg and Zeng 1992) were developed for less than 360°, non-circular detector orbit acquisitions. The algorithms require that data are sampled over an angular range of 180° plus the fan or cone angle plus an additional small angle. For these conditions the algorithms accurately reconstruct fan-beam geometry but can only accurately reconstruct the central plane of a cone-beam projection acquisition since the reconstruction of the midplane is equivalent to the reconstruction of a fan-beam data set. Therefore, short-scan cone-beam reconstructions of the midplane should be equivalent to full-scan cone-beam and parallel reconstructions if for the short-scan acquisitions the central slice is sampled over a sufficient angular range for fan-beam projections.

The computer-generated phantom in figure 11 is used here to analyse the short-scan cone-beam reconstruction algorithm (Gullberg and Zeng 1992). Simulations in figure 12 compare reconstructions of parallel projections (128 projections over 360°) with reconstructions of full-scan circular (128 projections), short-scan circular (83 projections) and short-scan non-circular (83 projections) cone-beam projection data sets. The projections of the source distribution were formed assuming no photon attenuation, thus focusing on the differences between short- and full-scan reconstructions. The parallel reconstruction is used as a standard for evaluating the reconstruction artifacts associated with single planar orbit cone-beam tomography.

Figure 11. The ellipsoidal heart-lung simulation phantom. The following source densities, \(\rho\), were used: 1 (ventricular wall), \(\rho = 1\); 2 (intraventricular blood pool), \(\rho = 0.2\); 3 (lungs), \(\rho = 0.2\); 4 (tissue), \(\rho = 0.2\). The unit voxel equals 0.6 cm. A defect (\(\rho = 0\)) is illustrated in the wall of the ventricle.

From the profiles, one sees that the reconstructions of the midplane (transverse slice 32) are equivalent, as one would expect, for all reconstructions. The cone-beam reconstructions of transaxial slices off the central slice are approximations due to the nature of the incomplete data sampling. However, one observes from the profiles in figure 12 that for a short distance off the midplane, the cone-beam full- and short-scan
Figure 12. Reconstructions of unattenuated projections of the simulated heart-lung phantom shown in figure 11. Profiles compare the quantitative accuracy between parallel geometry, cone-beam circular 'full-scan', circular 'short-scan', and non-circular 'short-scan' reconstructions.
reconstructions appear to be equivalent with the parallel reconstruction. Differences become more apparent for slices further from the midplane.

These differences are better seen in coronal views such as the coronal slice shown in figure 12 which is a distance of 10.5 voxels anterior to a central coronal slice. The profile for the parallel reconstruction has a higher amplitude than the profiles for the full- and short-scan cone-beam reconstructions. If we assume that the parallel reconstruction is accurate, one would infer that the differences in the profiles are due to the insufficient sampling of the cone-beam projections. Differences between short- and full-scan reconstructions are more apparent in background tissue where low-amplitude artifacts are seen extending horizontally near the superior and inferior extensions of the heart. Also, the cone-beam reconstructions show some additional low amplitude artifacts extending vertically along the axis at the periphery of the simulated background distribution. This is best appreciated in the short-scan reconstructions which exhibit a broadening of the profile at the boundary as compared with the full-scan and parallel reconstructions. It appears that the combination of the approximation of the cone-beam reconstruction algorithm for non-central slice voxels and the short-scan data set together result in artifacts that are seen in short-scan cone-beam reconstructions but are not seen in full-scan reconstructions.

3.3. Truncated projection correction algorithms

The magnification of a converging collimator can truncate both transmission and emission projection data. In some emission projections, the magnification can place the radiopharmaceutical distribution in the tissue surrounding the organ of interest outside the field-of-view. The application of the reconstruction filter to these truncated projections produces spikes at the truncation edge in the filtered projections (figure

![Figure 13](image_url)

Figure 13. A truncated projection profile of one sampled projection is shown (---). After filtering with a ramp filter the profile (-----) has a spike at the truncation edge. To eliminate the ring artifact in the reconstruction caused by the spike, the filtered projections are multiplied by a window function (---) to give the result shown with the full curve which is backprojected to form the reconstruction.
Convergent beam tomography

13), resulting in ring artifacts in the reconstruction (figure 14(a)). Clipping the truncation spikes after filtering and before backprojecting the filtered projections works well to eliminate the reconstruction ring artifacts (figure 14(b)) if, such as in cardiac SPECT, only the background distribution is truncated and is low compared to the organ of interest (Zeng et al 1990).

For transmission computed tomography (Malko et al 1986, Tsui et al 1989a, Manglos et al 1990), which is used to obtain an attenuation map for iterative attenuation-correction reconstruction algorithms (see section 3.4), the limited field-of-view of most SPECT detector systems makes it difficult to acquire transmission data of the thorax with converging collimators without truncating the projections in most of the projection views (Manglos et al 1990). Our approach (Tung et al 1991b) has been to reconstruct the attenuation distribution using an iterative EM reconstruction algorithm (Lang and Carson 1984). The iterative algorithm solves an underdetermined system of linear equations corresponding to only those projections that are measured. The attenuation factors which are exponentials of partial line integrals of the attenuation distribution are calculated from the attenuation distribution reconstructed from the truncated transmission projections.

Simulations in figure 15 show that even though the transmission image is distorted at the boundary the attenuation correction method produces quantitative emission reconstructions. Our hypothesis is that even though the transmission image is significantly distorted, the attenuation factors are measured accurately enough for those attenuation factors that have the greatest influence upon the emission measurements and, thus, are useful in producing quantitative emission reconstructions. The reconstruction of the truncated projections into an array larger than the size of the non-truncated portion of the projection, with the constraint that the support be equal to the outer boundary of the object, significantly improves the image quality of the transmission image and gives the best accuracy in the attenuation-correction reconstruction. However, we have seen from simulations that in general this is only somewhat

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**Figure 14.** Reconstructions of truncated projections (a) before and (b) after applying the window function illustrated in figure 13 to eliminate the truncation spikes. The transaxial slices of a $^{201}$TI cardiac patient study were reconstructed from a 'short-scan' data set of 76 projections obtained using the 50 cm focal length cone-beam collimator on the SX-300 SPECT system.
Figure 15. Simulation of attenuation correction using a truncated transmission scan. Transmission images were reconstructed from a set of 120 fan-beam projections sampled over 360° using 25 iterations of the EM algorithm. The focal length of the fan-beam geometry was 70 pixels (1 pixel = 0.72 cm) and the elliptical phantom had an outer boundary with major and minor axes of 50 and 30 pixels respectively. The transmission and emission projections were truncated symmetrically by 20 pixels on each side of each projection. Firstly, the truncated transmission projections were reconstructed into a 24 x 24 array equal to the dimensions of the non-truncated projections. Secondly, the truncated projections were reconstructed into a 64 x 64 array. Thirdly, the truncated transmission projections were reconstructed into a 64 x 64 array but the reconstruction was restricted to an elliptical support equal to the outer boundary of the phantom. The emission images were reconstructed using 25 iterations of the EM algorithm. The attenuation factors for the emission reconstruction were calculated from each of the three reconstructed attenuation distributions for the cases described above.
better than using the transmission reconstruction without support even if it is significantly distorted and in particular from the simulations in figure 15 it is difficult to see any difference.

3.4. Iterative attenuation and geometric point response correction algorithms

Iterative reconstruction algorithms are computationally more intensive, especially for cone-beam geometry (Manglos 1989, Manglos et al 1989a), than filtered backprojection reconstruction algorithms but have the advantage of being able to accurately compensate for physical factors of attenuation (Gullberg et al 1989, Manglos et al 1991) and a spatially varying geometric point response (Floyd et al 1986b, Tsui et al 1988, Formiconi et al 1989, Zeng et al 1991, Zeng and Gullberg 1992b, Liang et al 1992). In particular, it is recognized that photon attenuation within the body is a major factor contributing to the quantitative inaccuracy using SPECT to measure the in vivo distribution of radioactivity (Gullberg et al 1982, Lewis et al 1982, Webb et al 1983, Malko et al 1986, Tsui et al 1989a). It has been shown that iterative attenuation correction algorithms which model the attenuation process using a measured attenuation distribution of the thorax more accurately quantify the source distribution in the heart (Tsui et al 1989a) than the more efficient filtered backprojection algorithms which use pre-processing or post-processing attenuation correction techniques based upon either an assumed constant attenuator or a measurement of the variable attenuator. The disadvantage with single-detector SPECT systems has been the necessity to either acquire simultaneous transmission and emission data (Bailey et al 1987, Frey et al 1992), thus contaminating either transmission or emission data because of cross-talk between transmission and emission energy windows, or acquire separate emission and transmission scans thus doubling the patient imaging time.

The new multi-detector systems (see § 2.1) offer the ability to simultaneously acquire transmission and emission data (Tung et al 1991a), thus making it feasible to acquire the data necessary for variable attenuation correction in an acceptable clinical imaging time. A transmission line source is positioned opposite one of the detectors (figure 4, detector 3) of a multi-detector, fan-beam collimated SPECT system. Transmission and emission data are acquired in one detector (3) while the other two detectors (1 and 2) simultaneously acquire emission data. Cross-talk from one source into the energy window of the other source for the detector acquiring simultaneous transmission and emission data is removed based upon separate measurements of scatter; one using only the transmission source and one using only the emission source. For the data obtained from the detector acquiring simultaneous transmission and emission data, the fraction of cross-talk is removed from the projections before reconstruction. For the detectors acquiring only emission data the cross-talk is less but is also measured and subtracted out. If 99mTc is used as the transmission source and 201Tl as the emission source, an estimate of the high-energy 201Tl photons that will appear in the 99mTc window of the detector for the simultaneous transmission and emission data is obtained from measurements of the 99mTc window of the other two detectors during the scan. If 99mTc is used for both emission and transmission sources, the transmission data are obtained from the detector obtaining the simultaneous transmission and emission data by subtracting estimates of the emission data obtained from the other two detectors.

A simultaneous transmission and emission experiment was performed using the Jaszczak elliptical phantom with cardiac insert shown in figure 16(a). The transmission
Figure 16. (a) Experimental set-up of the simultaneous emission and transmission phantom experiment on the PRISM 3000. The transmission line source filled with $^{99}$Tc is attached to the detector so that it remains at the focal line as the detector follows a non-circular orbit. The heart insert and background of the elliptical Jaszczak phantom were filled with $^{201}$Tl. (b) Results of short axis reconstructions without and with attenuation correction using 25 iterations of EM algorithm and attenuation factors calculated from the reconstructed attenuation distribution (not shown). The emission data from detectors 1 and 2 in figure 4 were used in the reconstruction.
Convergent beam tomography

line source was filled with $7.4 \times 10^8$ Bq (20 mCi) of $^{99m}$Tc, the wall and chamber of the cardiac insert were filled with $4.44 \times 10^6$ Bq (120 μCi) and $1.14 \times 10^7$ Bq (309 μCi) of $^{201}$TI respectively and the background in the outer elliptical chamber was filled with $1.14 \times 10^6$ Bq (309 μCi) of $^{201}$TI. Both the transmission (detector 3) and emission (detectors 1 and 2) projection data were reconstructed from 120 projections sampled over $360^\circ$ using 30 iterations of the EM algorithm. The results in figure 16(b) demonstrate an improvement in the uniform distribution of the in vivo radionuclide concentration for the attenuation-corrected reconstruction.

In addition to attenuation it is also recognized that the spatially varying point response of a gamma camera due to the collimator geometric response and scatter results in loss of contrast and reconstruction artifacts that are seen as shape distortions and density non-uniformity. Models for photon attenuation (Gullberg et al 1989, Manglos et al 1991), the three-dimensional spatially varying geometric point response (Tsui and Gullberg 1990) and scatter (Liang et al 1992) are incorporated into projection and backprojection operations that are used in iterative reconstruction algorithms to correct for these physical effects. In one approach, the space varying point response can be modelled as a three-dimensional ‘inverse cone’ of rays emanating from each detector bin (Zeng et al 1991), which transforms even parallel tomography into a cone-beam reconstruction problem. The projection and backprojection operations sum along each ray weighting the contribution to the sum by the geometric response and attenuation factors that are calculated during the projection and backprojection summation along each ray. Another approach (Zeng and Gullberg 1992b, Liang et al 1992) which is computationally more efficient is to use the shift invariant property of the collimator geometric response at a fixed distance from the detector. In 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 result of correcting for the spatially varying geometric point response in a cone-beam tomography study of a patient heart is shown in figure 17. The corrected images show a slight narrowing in the wall thickness of the left ventricle. In the projection and backprojection operations the three-dimensional ‘inverse-cone’ structure was used to model only the geometric response of the collimator and no attenuation or scatter correction was implemented. Therefore, it is difficult to interpret the results considering the complication of potential reconstructed attenuation artifacts being present. To evaluate only the geometric response correction, a phantom study was performed using the Jaszczak heart phantom without filling the outer cylindrical portion with water. This minimized the attenuation in the projections of the source distributed in the cardiac insert, thus allowing us to focus upon and to evaluate the point response correction. The projections were obtained using a fan-beam collimator (50 cm focal length) on the PRISM 3000 SPECT system. The results in figure 18 show significant narrowing of the simulated heart wall with the geometric response correction.

3.5. Non-planar cone-beam reconstruction algorithms

As discussed in section 2.3, in cone-beam SPECT it is common practice to acquire scans tranversing a single circular or non-circular planar orbit (figure 8(a)) and reconstruct the projection data using a Feldkamp-type reconstruction algorithm. These orbits do
Figure 17. Correction for the spatially varying geometric point response in a cone-beam tomography study of a patient heart. An iterative EM algorithm was used to reconstruct 76 projections sampled over an angular range of approximately 214°. The projections were acquired on the SX-300 SPECT system using the 50 cm focal length cone-beam collimator. The images were reconstructed without (on the left) and with (on the right) modelling for the geometric response in the projection and backprojection operations. Modelling of photon attenuation and scatter was not included.

not satisfy the data sufficiency condition since one can find planes that pass through the reconstructed region-of-interest, parallel to, but not intersecting, the plane of the orbit. Therefore, reconstruction artifacts can result and these artifacts due to asymmetric reconstructed point responses have to be weighed against the advantage gained with increased sensitivity of cone-beam over parallel and fan-beam collimation.

Recognizing the potential problems with insufficient sampling of cone-beam projections, research is presently pursuing the use of non-planar orbit trajectories such as those shown in figures 8, 9 and 10 that would give sufficient angular sampling and could be implemented with minor modifications to present SPECT systems. The cone-beam reconstruction algorithm developed by Grangeat (1987) can be used to reconstruct cone-beam projections acquired from almost all orbit trajectories. However, the one drawback of the approach is that it involves the sorting (or rebinning) of the cone-beam data. We emphasize that even if the data sufficiency condition is satisfied, an efficient convolution backprojection reconstruction algorithm is not guaranteed. Two orbits for which efficient reconstruction algorithms have been derived are dual orthogonal (Clack et al 1991) and circle-and-line (figure 8(b)) orbits. Note that the orbit in figure 8(c) has one complete orbit and one orthogonal half orbit which in practice is necessary because one cannot pass the detector through the body. An algorithm for this orbit has not been developed. The algorithm developed by Clack et al (1991) was developed for two complete orthogonal orbits and approximates the cone-beam projections as Radon planar integrals. The circle-and-line algorithm (Zeng and Gullberg 1992a) uses a modification of Smith's algorithm (Smith 1990). The development of efficient filtered backprojection reconstruction algorithms for other useful detector trajectories, such
Convergent beam tomography

Figure 18. A phantom study was performed to evaluate the ability to correct for a space varying geometric response of a fan-beam collimator (50 cm focal length) on PRISM. Jaszczak heart phantom without water attenuator was used. Results of filtered backprojection algorithm (FBP), EM without and with geometric response correction. The results were reconstructed from 120 projections over an angular range of 360°. Modelling of photon attenuation and scatter was not included.

as the helical orbit (figure 8(d)), is presently an area of research which has the potential of significant future benefits.

Results in figure 19 compare the Feldkamp and iterative EM reconstruction of cone-beam projections acquired using a planar circular orbit, with the reconstruction of cone-beam projections acquired using dual orthogonal and circle-and-line orbits. The sagittal reconstructions of the Defrise phantom are superior for the orbits which satisfy the data sufficiency condition. However, notice that the iterative EM reconstruction does show an improvement over the Feldkamp reconstruction for a planar circular orbit.

3.6. Geometrical parameter estimation algorithms

In addition to the need for special reconstruction algorithms for converging geometries, it is important for good tomographic image quality to be able to measure precisely the geometric parameters of the physical imaging system so as to accurately orient in space collimator, camera and rotating gantry relative to the centre of rotation. A method was developed to estimate the geometrical parameters for fan-beam (Gullberg et al 1987) and cone-beam (Gullberg et al 1990) detector geometries from measured coordinates of the centroid of a projected point source sampled over 360°. Non-linear expressions
are derived for the coordinates of the centroids in terms of the geometrical parameters. The Marquardt algorithm is used to determine the geometric parameters by fitting the measured centroid data to the non-linear expressions of the geometric parameters. The method is sensitive to systematic errors and is also useful to evaluate the accuracy of the collimator construction.

3.7. Variance reconstruction algorithm

An important aspect in evaluating the clinical efficacy of cone-beam tomography is to determine the specificity and sensitivity of the technique which is most accurately ascertained through very time consuming observer studies. As an initial attempt to at least evaluate contrast resolution, an algorithm (Hahn et al 1988) was developed to estimate the variance of the estimated intensity for each pixel reconstructed using the Feldkamp algorithm. The estimated variance measures precession and thus is useful to evaluate the minimal detectable contrast resolution. The approach developed by Hahn et al (1988) can also be used to obtain covariance estimates which are useful to evaluate the texture in the cone-beam reconstructions. We have also found that the spatial distribution of variance estimates is useful to evaluate the asymmetry of the reconstructed point response for single planar orbit cone-beam tomography.
4. Results

The development of hardware and algorithms has made possible the clinical evaluation of converging-beam tomography. Preliminary results of phantom and patient studies are presented here. These studies were performed using the collimators listed in table 1. The data were acquired from single planar orbits on the SX-300 and PRISM 3000 SPECT systems (Ohio Imaging of Picker International, Bedford Heights, Ohio).

Table 1. Collimators used in the phantom and patient studies. The hole length is for the central hole of the converging collimators, the hole diameter is the measurement at the centre of the tapered hole, and the focal length is measured from the front face of the collimator. The tabulated resolution (FWHM) is the total system resolution (intrinsic plus collimator).

<table>
<thead>
<tr>
<th>Collimator Type</th>
<th>Resolution at 10 cm (mm)</th>
<th>Sensitivity (counts min(^{-1}) (\mu\text{Ci}))</th>
<th>Hole diameter (mm)</th>
<th>Hole length (cm)</th>
<th>Focal length (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cone beam (SX-300)</td>
<td>6.0</td>
<td>270</td>
<td>1.89</td>
<td>4.0</td>
<td>50</td>
</tr>
<tr>
<td>Cone beam (SX-300)</td>
<td>6.7</td>
<td>252</td>
<td>1.55</td>
<td>3.3</td>
<td>60</td>
</tr>
<tr>
<td>Fan beam (SX-300)</td>
<td>11.4</td>
<td>243</td>
<td>6.0</td>
<td>13.0</td>
<td>58</td>
</tr>
<tr>
<td>Fan beam (PRISM 3000)</td>
<td>7.8</td>
<td>300</td>
<td>1.4</td>
<td>2.7</td>
<td>50</td>
</tr>
<tr>
<td>Parallel (SX-300)</td>
<td>9.1</td>
<td>350</td>
<td>1.6</td>
<td>2.54</td>
<td>(\infty)</td>
</tr>
<tr>
<td>Parallel (PRISM 3000)</td>
<td>7.8</td>
<td>245</td>
<td>1.4</td>
<td>2.7</td>
<td>(\infty)</td>
</tr>
</tbody>
</table>

\(1 \text{ Ci} = 3.7 \times 10^{10} \text{ Bq.}\)

4.1. Phantom study

A study was performed on the SX-300 SPECT system using the 20 cm circular Jaszczak heart phantom. Experimental results in figure 20 show parallel and cone-beam reconstructions of transaxial and short axis slices through the cardiac insert. The LEGP parallel hole collimator was used for the parallel study and the 60 cm focal length cone-beam collimator was used for the cone-beam study which differs from the studies by Gullberg et al (1991a) where a 50 cm focal length cone-beam collimator was used. The cone-beam results were obtained using the full-scan cone-beam reconstruction algorithm (Feldkamp et al 1984) to reconstruct 128 projections (=2.8° between projections) sampled over 360°. There was no pre- or post-filtering applied for noise suppression. The central slice of the cone-beam geometry which is equivalent to a fan-beam projection data set was summed over 128 projection angles to give 56 713 counts. The parallel-hole collimator results were reconstructed from 128 projections sampled over 360° of which a typical slice through the heart has a total of 126 597 counts summed over 128 projections. Also, for the parallel reconstructions no filtering for noise was used.

In comparing the results, we attempted to select anatomically the same transaxial and short axis slices for each study. The improved resolution for the cone-beam collimator is evidenced by the thinner region of activity. The wall thickness was measured for the short axis slice shown in figure 20 to be 1.46 cm for the cone-beam
results and 2.36 cm for the LEGP parallel-hole collimator results which compares to a known thickness of 1 cm. For the LEGP parallel hole collimator, the %RMS error was 14.62% ($\sigma = 28.19 \pm 2.81$, $\mu = 192.88 \pm 3.99$) as compared to 18.12% ($\sigma = 32.96 \pm 2.85$, $\mu = 181.88 \pm 4.03$) for the cone-beam results.

The difference in reconstructed resolution between the cone-beam and LEGP parallel-hole collimators correlates with a system resolution for the cone-beam collimator of 6.7 mm FWHM at 10 cm and for the LEGP parallel-hole collimator of 9.1 mm FWHM at 10 cm. The difference in the %RMS error is attributed to differences in sensitivity (sensitivity for the LEGP parallel-hole collimator of 1.39 times that of the cone-beam collimator) and differences in the slice thickness (0.63 cm for parallel and 0.45 cm for cone beam).

4.2. Patient studies

Converging collimation has been applied in several patient studies of heart (Gullberg et al 1991a) and brain. Here we show one heart study and one brain study.

4.2.1. Heart. Figure 21 shows a representative scan from a patient study demonstrating improved resolution of cone-beam cardiac SPECT. The cone-beam study was performed using the 60 cm focal length cone-beam collimator on the SX-300 and the parallel study was performed using the LEHR parallel-hole collimator on the PRISM. The patient was exercised and at the peak of stress was injected with $1.11 \times 10^6$ Bq (3 mCi)
Figure 21. Cardiac patient study. A comparison is shown between parallel and cone-beam reconstructions of cardiac transaxial and short axis slices. The parallel study used the LEHR parallel hole on the PRISM SPECT system to obtain 60 projections over 180° of approximately 125,034 total counts for a central transaxial slice. The cone-beam study used the 60 cm focal length cone-beam collimator on the SX-300 SPECT system to obtain 64 projections over 180° of approximately 49,116 total counts for the central cone-beam transaxial slice.

The better resolution in the cone-beam studies is seen with the thinner ventricular wall. The wall thickness is 1.64 cm for the cone-beam results and 2.27 cm for the LEHR parallel-hole collimator results. The study was done with a 60 cm focal length collimator which we designed especially for the heart. It is felt that a 60 cm or even longer focal length cone-beam collimator would have fewer problems with truncating the heart than the 50 cm focal length cone-beam collimator used by Gullberg et al (1991a).

4.2.2. Brain. Figure 22 shows a comparison of fan-beam and cone-beam tomography study of the brain performed on the SX-300 SPECT system. The cone-beam study was performed using the 50 cm focal length cone-beam collimator and the fan-beam study was performed using the 58 cm long-bore fan beam collimator (Tsui et al 1986). The patient was injected with $9.25 \times 10^8$ Bq (25 mCi) of $^{99m}$Tc$^-\text{HMPAO}$ and 128 projects were acquired for both studies. The cone-beam study was performed using a tilted cone-beam acquisition shown in figure 3 (Manglos 1989, Manglos et al 1989b) allowing the detector of $^{201}$Tl. The cone-beam results were obtained using the short-scan cone-beam reconstruction algorithm (Gullberg and Zeng 1992) to reconstruct 64 projections ($\approx 2.8°$ between projections) sampled over 180°. (Because of technical problems the angular sampling range was less than the minimum required for the short-scan algorithm.) The central slice of the cone-beam geometry which is equivalent to a fan-beam projection data set was summed over 64 projection angles to give 49,116 counts. The parallel-hole collimator results were reconstructed from 60 projections of which a typical slice through the heart had a total of 125,034 counts summed over 60 projections. Before reconstruction, both the cone-beam and parallel projections were filtered with a two-dimensional Metz filter.
to stay close to the head but still clear the patient shoulders. The long-bore design of the fan-beam collimator accomplished the same purpose without having to tilt the detector.

The cone-beam results show better resolution of structures in the brain than obtained with the fan beam collimator. However, we see image elongation along the axis of rotation near the top of the head in coronal and sagittal views due to the insufficient
convergent beam tomography

The tilted detector acquisition makes this problem even worse because the tilt of the detector places the top of the head further from the central plane of the cone-beam collimator. A non-stationary filter has been developed by Cao and Tsui (1992) which improves the slice-to-slice cross talk and image elongation.

5. Summary

Convergent-beam geometry provides improved sensitivity and improved resolution for both emission and transmission computed tomography. The development of collimators and algorithms has made possible the evaluation of this technology and the preliminary results of imaging brain and heart are showing significant potential for the clinical application of SPECT with convergent-beam geometry. We have found that a single-detector SPECT system using a high-resolution cone-beam collimator with tilted detector acquisition can obtain high-resolution transaxial images of the brain. However, with single planar orbit acquisition, we see image elongation and slice-to-slice cross-talk due to insufficient cone-beam data acquisition. This elongation and slice-to-slice crosstalk does not seem to be as much a problem with cone tomography of the heart since the heart does not extend axially as far as the brain from the central plane of the cone-beam geometry. However, detector systems and algorithms for non-planar orbit cone-beam tomography are being developed which eliminate these reconstruction artifacts associated with insufficient acquisition of cone-beam projections.

The development of convergent-beam geometry in SPECT requires the integration of hardware and software through the optimization of reconstruction algorithms, collimator design, orbit specifications and calibration techniques to obtain the best image quality and to minimize the effects of photon attenuation, insufficient projection sampling and data truncation. The challenge in using convergent-beam geometry for SPECT is to continue to improve computing hardware in order to process, in a reasonable clinical time, images that are reconstructed using algorithms that better model the physics of the imaging detection process and are useful for complicated detector geometries such as cone beam which gives improved geometric sensitivity. This technology has the potential to improve clinical diagnosis.

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Résumé

Mise au point sur la tomographie par faisceau convergent en tomographie d'émission à photon unique.

L'apport clinique des géométries convergentes en tomographie d'émission à photon unique (SPECT) fait actuellement l'objet d'études approfondies. Les collimateurs sont utilisés sur les caméras à scintillations tournantes 'grand champ' astigmate de type: en éventail, conique, sténopé, afin d'augmenter la sensibilité, aussi bien en tomographie d'émission que de transmission, pour des petits organes tels que la thyroïde, le
carreau ou le coeur. Avec les nouveaux systèmes SPECT multi-têtes, la géométrie convergente offre la possibilité d’obtenir simultanément les données d’émission et de transmission nécessaires à la quantification de la fixation d’une substance radiopharmaceutique dans le coeur. En SPECT, le développement des faisceaux à géométrie convergente nécessite l’intégration de logiciels et de circuits électroniques dans les systèmes existants. Du point de vue du matériel, le système de détection optimum pour la tomographie à faisceau conique est un système qui satisfait à la condition suffisante selon laquelle les trajectoires halées interceptent n’importe quel plan passant à travers la zone reconstruite. Cependant, la plupart des algorithmes ont été développés dans le cas de données insuffisantes, acquises suivant des orbites à un seul plan. Le développement de ces algorithmes a rendu possible l’évaluation préliminaire de cette technologie, et les images de carreau et de coeur ont montré la potentialité de la tomographie à faisceau conique pour les applications cliniques. Actuellement, des activités de recherche se poursuivent de façon significative dans le domaine des algorithmes adaptés au traitement des données acquises respectant ‘la condition de données suffisantes’ et pouvant être facilement et économiquement implantés sur les appareils SPECT utilisés en clinique.

Zusammenfassung

Überblick über die Konvergenzstrahl-Tomographie in der Einzel-Photon-Emissionscomputertomographie.


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