INSTRUMENTATION

Single-Photon Emission Tomography with a 12-Pinhole Collimator

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To assess the advantages of more complete angular sampling and of more views in the tomographic reconstruction process, tomographic imaging with a 12-pinhole (12PH) collimator has been compared with 7-pinhole tomography (7PH). The 12PH system gives a 50% increase in sensitivity but resolution degrades more rapidly with depth. The 7PH and 12PH systems provide similar accuracy of detection of lesions in a myocardial ring phantom. The 7PH images, however, demonstrated more noise and "ripple" artifacts. The 12PH system offers a larger reconstruction volume and generates fewer artifacts when the collimator is misaligned with the myocardial long-axis, thus making patient positioning less critical than with 7PH. A disadvantage is that individual views are minified by the 12PH collimator, and a 256 × 256 image matrix should be used during image acquisition to limit digital sampling errors.

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Seven-pinhole (7PH) tomography (*1,2*) was designed for quantification of regional myocardial Tl-201 perfusion and for other applications (*3*) in cardiovascular nuclear medicine. A 7PH collimator, attached to a scintillation camera, allows multiple independent views to be acquired simultaneously with different angular orientations without the use of moving collimators or detectors. Tomographic capabilities can be added to conventional scintigraphic systems at modest expense using computer facilities commonly available in many nuclear medicine departments (*4*). Rizi et al. (*5*) found that 7PH tomography represents a significant advance over planar Tl-201 imaging as a method of quantifying myocardial perfusion abnormalities.

Limited-angle tomographic systems, such as 7PH tomography, have recognized limitations. Careful patient positioning (*5*) and excellent intrinsic camera performance (*6*) are required to prevent generation of artifacts in the reconstructed tomograms. In addition, 7PH tomographic systems suffer from degradations in spatial resolution and sensitivity with increasing depth (*7*). Some limitations of 7PH tomography might be reduced by more complete angular sampling of the radionuclide distribution under study. To test this hypothesis, a twelve-pinhole (12PH) tomographic system has been developed and tested to investigate the impact of this collimator design on tomographic image acquisition and processing. Intuitively, increasing the number of views should increase photon sensitivity or allow decreased pinhole size to improve spatial resolution while maintaining constant sensitivity. More complete angular sampling also should decrease the criticality of patient positioning (*8*). However, increasing the number of views increases artifacts associated with the minification of the images projected onto the detector surface (Fig. 1) and the increased angle of incidence of photons at the periphery of the detector surface (*9*). The original 7PH collimator was designed for a camera with a spatial resolution of 4.5 mm FWHM. Latest large-field camera designs achieve 3.5 mm FWHM. This improvement indicates that the diameter of individual views can be decreased from 12.5 cm to about 10 cm without loss of resolution in comparison with the original tomographic

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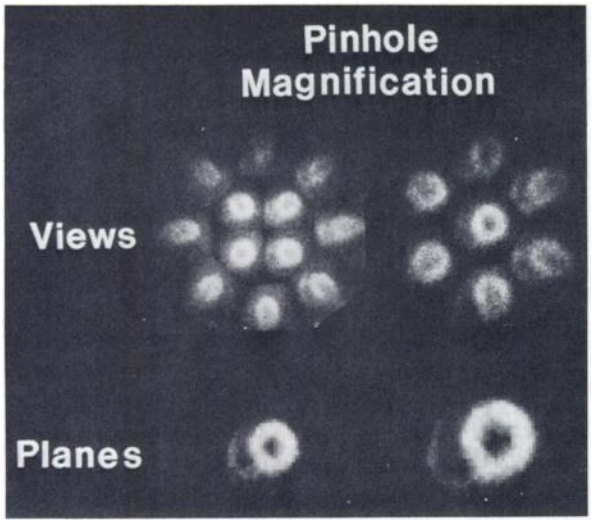


FIG. 1. Acquired images (top) and tomographic planes (bottom) of Tl-201 myocardial perfusion study using 12PH (left) and 7PH (right) collimators. 12PH study shows effect of image minification in both acquired images and tomographic reconstructions.

system. This corresponds to an increase in the number of views from 7 to 12. Additional increases in the number of views would cause additional minification and would limit the quality of the tomograms owing to the finite resolution of the detector.

To assess the cumulative effect of changes in collimator design, measurement of spatial resolution from point sources and reconstructions of myocardial phantoms were performed and evaluated. These results are discussed in terms of the advantages that can be obtained from a redesign of the collimator and how these features might affect the use of N-pinhole tomography in a clinical setting.

MATERIALS AND METHODS

Collimators. The 7PH collimator was obtained commercially* and has been described previously (*4*). The 12PH collimator is similar to the 7PH collimator (Figs. 1 and 2) except that the former has four central views surrounded by eight peripheral views. The pinholes are spaced at regular angular increments around circles of 3.56 cm and 9.14 cm radius, centered on the pinhole plate. Interchangeable plates with pinhole diameters of 4.0 and 6.5 mm were used with the 12PH collimator. The 7PH collimator had 6.5-mm pinholes.

The increase in the number of views was accomplished by decreasing the distance between the pinhole plate and the detector surface from 12.7 to 7.6 cm. This change decreased the size of the individual projected views while increasing sampling angles and widening the field of view of the collimator (Fig. 2). The sampling angle can be specified as the angle between the line of sight through a pinhole to the point of interest and the axis perpendicular to the pinhole plate of the collimator. For a depth of 12 cm from the pinhole plate, the inner and outer pinholes of the 12PH collimator provide sampling angles of 16.5° and 37.3°, respectively. The 7PH collimator provides a 27.9° sampling angle. The common field viewed by all pinholes has a maximum diameter of 20.3 cm for the 12PH collimator compared with 12.7 cm for the 7PH collimator (Fig. 2).

Calibration sources. Flood and point calibration sources of Au-195 were obtained commercially and were used with a Au-195 myocardial phantom described below. Since the response of the scintigraphic system can change with energy (*9*), Tc-99m flood and point sources were imaged to calibrate Tc-99m point-source images acquired for determination of spatial resolution.

Point sources for spatial resolution measurements. For these measurements small drops (<1 mm diameter) of a pertechnetate solution were placed on acrylic plates and allowed to dry before imaging.

Myocardial phantom. The Au-195 myocardial phantom consists of one cylindrical endcap and 7 rings, each 1.8 cm thick. Four "normal" rings with uniform distributions of radioactivity have radial thicknesses of 1.0 cm and outer diameters of 6.0, 7.0, 8.0, and 9.0 cm. The cylindrical endcap has a diameter of 5.0 cm. The rings are stacked in the shape of a truncated cone to form the walls of the phantom, with the endcap placed on top to mimic the myocardial apex. Three rings with 90° sectors of 50% decreased radioactivity levels have the same diameters as the smaller three normal rings. These "abnormal" rings are substituted for the normal rings to simulate myocardial Tl-201 perfusion defects of different thicknesses at various depths.

Image acquisition. Images were acquired as 128 × 128 pixel, 16-bit matrices and stored in a computer interfaced to a gamma camera. Calibration flood and point-source images contained 5 million and 40,000 counts, respectively. Point-source images acquired for spatial resolution measurements contained 40,000 counts and myocardial phantom images contained 750,000 counts. Data were acquired at rates of 10,000 cps or less.

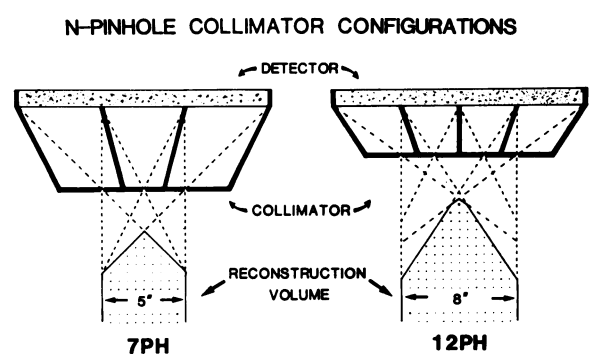


FIG. 2. Cross-sectional representation of 7PH and 12PH collimator configurations. 12PH collimator has decreased from distance pinhole plate to detector surface, resulting in additional views, minified images, and larger construction volume.

Image processing. Following acquisition, images were scaled linearly to a 128 × 128 pixel, 8-bit format and transferred to a PDP 11/34 computer† for processing. The tomographic algorithm used the "Interactive Data Language".‡ A single computer program was written to process either 7PH or 12PH images, so differences in the reconstructed tomograms are not due to differences in software and therefore can be attributed to differences in collimator design. Besides the capability to process either 7PH or 12PH images, the algorithm differs from the process described by LeFree (*6*) in that fractional rather than integral pixel shifts were performed during the reconstruction (Appendix). Use of fractional shifting is especially important for reconstruction of tomographic planes from the minified 12PH images because of the relative coarseness of integral pixel shifts. Fractional pixel shifting also allowed comparable spacing of the tomographic planes to be obtained from the two systems.

XY resolution. Resolution measurements were obtained from tomograms of single point sources. For measurements parallel to the detector surface ("xy resolution"), processing parameters were adjusted so that the central tomographic plane occurred at the depth of the point source. The peak value in the central plane of the point-source tomogram was interpolated from a second-degree polynomial fit to the three largest pixel contents along the direction of measurement. The half-maximum locations were interpolated linearly from pixel values closest to the actual half-maximum values.

Measurements were calibrated for distance by simultaneous imaging of two point sources and by division of their physical distance by the separation of their maxima in the tomograms. Distance calibration factors were obtained independently for the 7PH and 12PH systems, and for each depth of the point source.

Z-axis resolution measurements. For measurements of spatial resolution along the central axis perpendicular to the detector surface ("z resolution"), a minimum of six tomographic planes were reconstructed through the full-width-at-half-maximum (FWHM) region of each point source. The point-source profile in the z direction was extracted from the pixels in the center of the point-source reconstruction from successive tomographic planes. The peak value was interpolated from a second-degree polynomial fit to the three largest values in the profile. The plane-number positions of the half-maximum values were interpolated linearly from pixel values neighboring the exact half-maximum value.

The relationship between plane number and physical distance is not linear because the equal pixel shifts used in the reconstruction algorithm result in unequal spacings between the tomographic planes. To calibrate the plane positions for depth, seven point sources were placed at various depths along the perpendicular axis, were imaged separately, and tomograms were reconstructed. The known locations of these point sources were used to convert the measurements of z resolution from units of plane numbers into units of physical depth.

Myocardial phantom studies. Images of the myocardial phantom were acquired and tomograms were reconstructed. Characteristics of the tomograms were quantified using both conventional count profiles as well as "circumferential" count-profile analyses (*2*).

RESULTS

System sensitivity. The photon detection efficiency or sensitivity of the tomographic system was determined by dividing the counts detected by the time duration of image acquisitions with the Au-195 myocardial phantom. At a depth of 10 cm, the 7PH collimator provided a count sensitivity of 2000 cpm/μCi. For the same distance between pinhole plate and phantom, and for identical pinhole diameters, the 12PH collimator gave a 50% increase in sensitivity over the 7PH collimator. For both collimators, the sensitivity changed in direct proportion to the areas of the pinholes when the plates with different pinhole diameters were used.

Z resolution. For a constant pinhole diameter, the 12PH system had resolution characteristics comparable to those of the 7PH system only for points less than 10 cm from the pinhole plate (Fig. 3). The resolution of the 12PH system degraded more rapidly with depth than did that of the 7PH system. One can improve the spatial resolution of both systems by decreasing the diameters of the pinholes, with accompanying loss of sensitivity. This improvement is not in direct proportion to the pinhole diameters since, especially for smaller pinhole diameters, spatial resolution characteristics are limited by

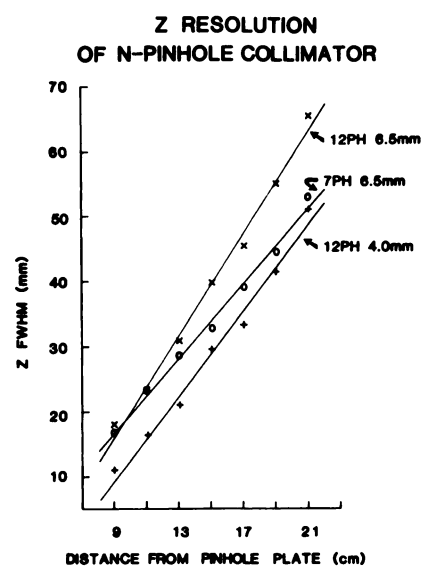


FIG. 3. Z-axis resolution for Tc-99m point sources as a function of distance from pinhole plate. Values are given for 12PH collimators with 6.5-mm and 4.0-mm pinhole diameters, and for 7PH collimator with 6.5-mm pinholes.

digital sampling, the intrinsic response of the detector, and the magnification characteristics of the collimator.

XY resolution. Characteristics similar to those for z resolution were observed in comparative measurements of xy spatial resolution. The resolutions of the 12PH system were best closest to the pinhole plate and degraded more rapidly with increasing distance than those of the 7Ph system (Fig. 4).

Myocardial phantom lesion detection. To quantify the accuracy of the tomographic reconstructions, tomograms of the myocardial phantom with a lesion were reconstructed and evaluated using circumferential profile analysis. The resulting circumferential profiles (Fig. 5), taken from the plane corresponding to the second ring with a lesion in the phantom, show good agreement with the known activity distribution both in angular extent and in the radioactivity decrease within the lesion.

Tomographic ripple. Tomograms of a phantom with no lesion were reconstructed to assess the presence of noise and "structured mottle" (*4*) in the images. Circumferential profile evaluations (Fig. 6) of the reconstructed tomogram of the second ring show, for all systems, fluctuations of the measured count densities about their mean levels. These fluctuations, or "tomographic ripple," are 1.5–2.5 times larger for the 7PH than for the 12PH system with identical pinhole diameters. The magnitude of the ripple artifact is related both to the number of views and to the resolution of the system. Comparison of results in Fig. 6 from the two 12PH systems shows that the resolution improvement with the 4.0-mm pinholes is accompanied by an increase in the ripple artifact. As the image is sharpened by the improved resolution, the ripple artifact becomes more prominent. The increase in ripple artifact with improved spatial resolution is one reason why pinhole diameters less than about 5.5 mm cannot be used with the 7PH collimator. However, the magnitude of the ripple artifact can be diminished by increasing the number of views.

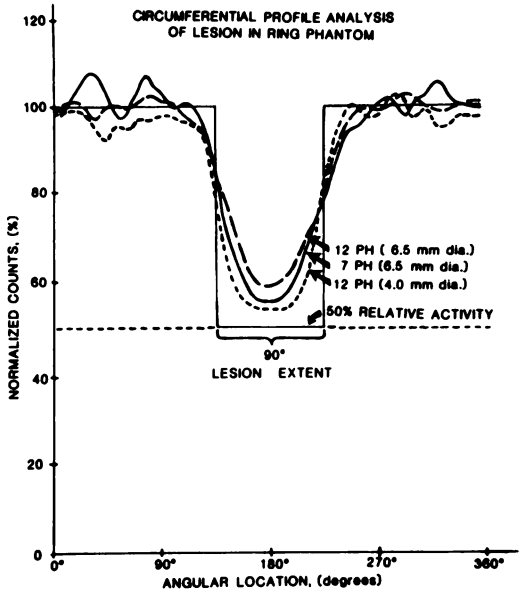


FIG. 5. Circumferential profiles from Au-195 myocardial phantom with 50% lesion. Profile is shown for known lesion characteristics (rectangular defect), and tomograms using 12PH collimators with 6.5-mm and 4.0-mm pinholes, and 7PH collimator with 6.5-mm pinholes. Mean count level in normal regions of all three circumferential profiles is taken as 100% and used to normalize individual profiles.

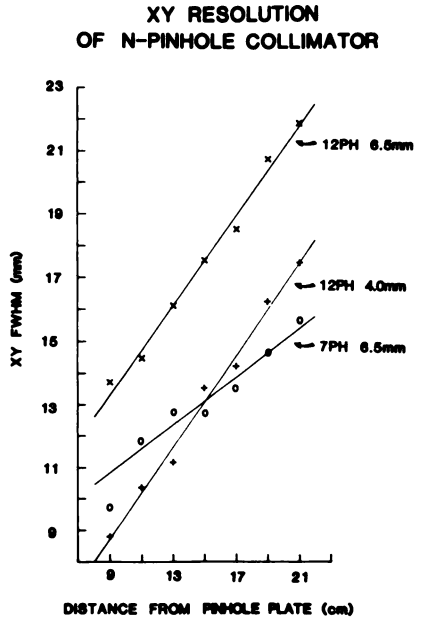
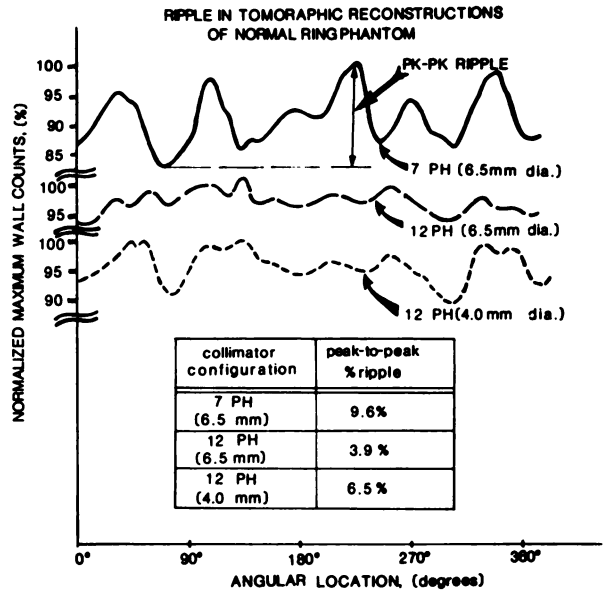


FIG. 4. XY spatial resolution for Tc-99m point sources as a function of distance from the pinhole plate. Values are for 12PH collimators wth 6.5-mm and 4.0-pinhole diameters, and for 7PH collimator with 6.5-mm pinholes.



collimator configuration	peak-to-peak % ripple
7 PH (6.5 mm)	9.6%
12 PH (6.5 mm)	3.9 %
12 PH (4.0 mm)	6.5 %

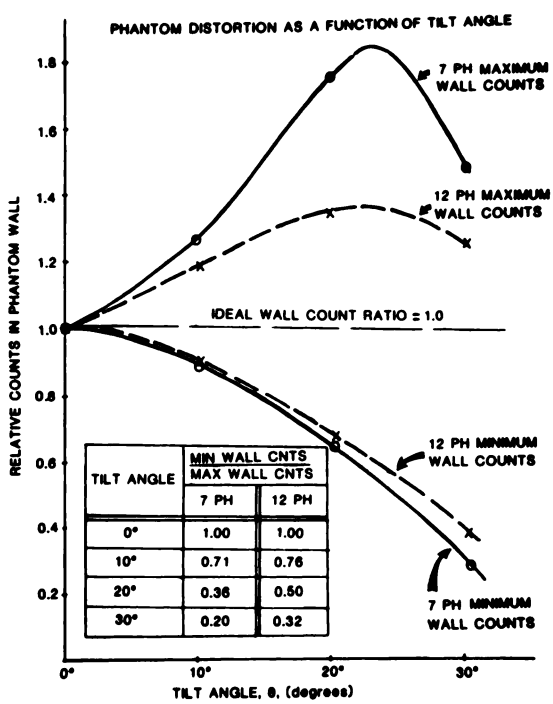
FIG. 6. Circumferential profiles from normal myocardial ring phantom using 12PH collimators with 6.5-mm and 4.0-mm pinholes and 7PH collimator with 6.5-mm pinholes. Ideal normal profile has 0% peak-to-peak ripple.

The curves in Fig. 6 were taken from a plane positioned 12 cm from the pinhole plate. At this point, the 7PH 6.5-mm collimator has a resolution of 12 mm FWHM, compared with a resolution of 11 mm FWHM for the 12PH 4.0-mm collimator. In spite of its improved resolution, the 12PH collimator also shows reduced ripple artifact (6.5% vs. 9.6%) due to the increased sampling obtained with this collimator.

Collimator misalignment artifacts. One of the limitations of pinhole and other limited-angle tomographic systems is the generation of artifacts when the collimator axis is misaligned with the long axis of the heart (*8*). To quantify this effect, a myocardial phantom without lesion was imaged with angles of 0°, 10°, 20°, and 30° between the collimator axis and the phantom axis of symmetry.

As one increases the angle of misalignment, one wall shows an increase in detected counts whereas the other shows a decrease. Count profiles were taken through the center of the tomograms along the line of greatest distortion. Figure 7 shows the maximum counts in the regions of the tomograms with the greatest increase and decrease in wall counts. The increase in maximum wall counts is a factor of 2 larger for the 7PH system than for the 12 PH system at the angle of maximum degradation. The decrease in distortion with collimator misalignment is a manifestation of the increased angular sampling of the 12PH collimator system (*8*).

Patient images. Tomographic reconstructions of a Tl-201 myocardial perfusion study from a normal patient are shown in Fig. 8. Our early impression is that the reconstructions obtained with the different systems are comparable in quality and in quantitative information. The 12PH system seems to yield clearer reconstructions of the myocardial apex, but otherwise shows degradation due to digital sampling and to the minification of images during acquisition. Implementation of a 256 × 256-pixel image matrix during acquisition should diminish this problem but would increase the time required for reconstruction of the tomograms.



TILT ANGLE	MIN WALL CNTS / MAX WALL CNTS	
	7 PH	12 PH
0°	1.00	1.00
10°	0.71	0.76
20°	0.36	0.50
30°	0.20	0.32

FIG. 7. Tomogram count distortion as function of collimator misalignment angle. Ideal wall-count ratio equals 1.0 for phantom with no defects.

DISCUSSION

The ability to reconstruct tomographic planes of a radionuclide distribution depends on the acquisition of multiple views of that object from different angles. The pinhole tomographic technique allows multiple views to be acquired simultaneously with a standard scintillation camera. The choice of only seven views in the original multiple-pinhole tomographic system was dictated by a large-field camera with instrinsic spatial resolution of 4.5 mm FWHM and with 1% spatial distortion in bar-phantom images. The benefit of increasing the number of views is limited by the minification of the images as more are acquired on a detector of fixed area. The design of a collimator for multiple-pinhole tomographic studies must account for the relative benefits of increased numbers of views versus limitations of intrinsic camera performance.

Recent developments have improved scintillation camera performance. Thinner crystals and photomultiplier tubes with smaller diameters (*10*) have resulted in better spatial resolution. Microprocessor technology has produced systems with virtually perfect uniformity and spatial linearity (*11*) as well as better energy resolution (*12*). These improvements in camera performance made it feasible to consider how tomographic image quality might be removed by redesigning the original 7PH collimator.

Spatial resolution is one apparent limitation of the 12PH system as compared with the 7PH system. For pinholes of the same diameter, the resolution in the 12PH system is worse and degrades more quickly with depth than in the 7PH system. This is the cumulative result of several factors, including intrinsic detector resolution and limitations of digital sampling. The magnification characteristics of the collimator is another consideration. For a pinhole collimator, the image size equals the object size when detector-to-pinhole and pinhole-to-object distances are equal. If the object-to-collimator distance is doubled, the image size is halved. The closer placement of the pinhole plate to the detector surface in the 12PH design causes the more rapid diminution of image size with depth; at an object depth of 15.2 cm, the image size was halved by the 12PH collimator and reduced by a factor of 1.2 using the 7PH. Since the resolution in the

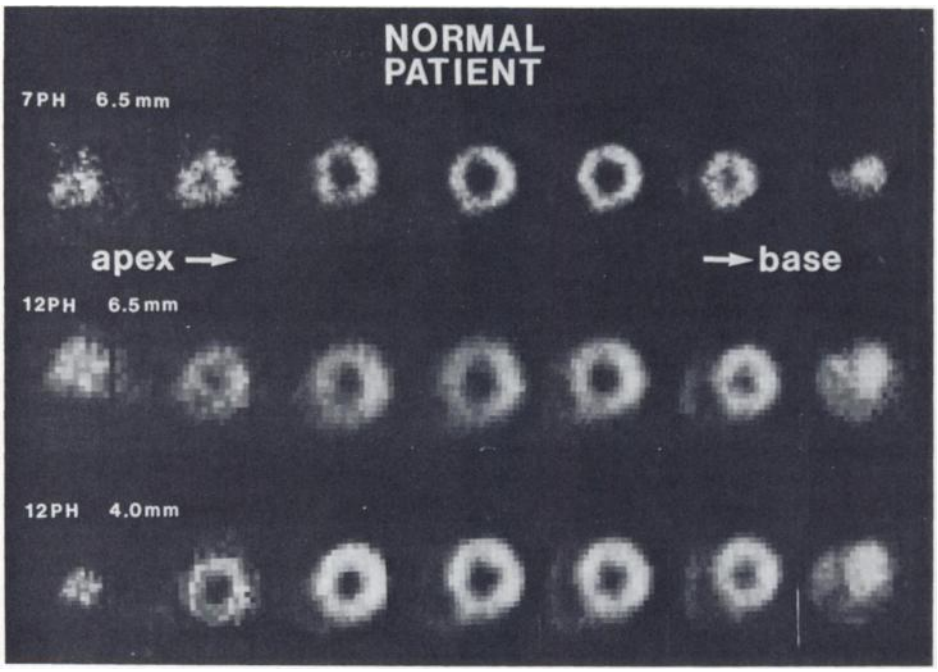


FIG. 8. Tomographic reconstruction of Tl-201 myocardial perfusion study from normal patients. 12PH tomograms show degradation due to digital sampling.

projected images is limited by the intrinsic resolution of the detector and the limitations of digital sampling, the degradation of system resolution should parallel the diminution of image size with depth. This more rapid degradation of resolution is one trade-off to the increase in view number that the 12PH collimator design affords.

Another factor is the elongation and position deviations of the point-spread response of a scintillation camera at large angles of photon incidence (*9*). These characteristics are more prominent with the 12PH collimator, owing to its larger sampling angles. For a longitudinal system, there is a limit to which one can increase sampling angles and the number of views before serious image degradation occurs. The results of the present study indicate that 12 views are probably the most that can be acquired on a single stationary large-field detector with currently available performance characteristics.

In a clinical setting, one probably would use a 12PH collimator with smaller pinhole diameters than those of 7PH collimators currently in use. The 12PH system provides more flexibility in improving spatial resolution by making this simple adjustment, since it demonstrates higher sensitivity and produces tomograms with less ripple. A 12PH system with smaller pinholes can be designed with sensitivity and resolution characteristics comparable with those of currently available 7PH systems, and would provide other advantages. For example, the 12PH collimator has a larger field of view. Also, the increase in angular sampling results in decreased artifact generation with collimator misalignment. Both of these factors serve to ease the requirements for patient positioning, and are obvious advantages of the 12PH system.

Another improvement could be made by increasing the image matrix to a 256 × 256-pixel format. Both phantom and patient images acquired with the 12PH system demonstrated obvious image degradation due to digital sampling. In order to take advantage of improved intrinsic resolution of currently available scintigraphic equipment, one must match digital sampling, as well as collimator geometries and pinholes diameters, to the intrinsic characteristics of those systems.

Final conclusions await comparative studies on patients. Trade-offs between system characteristics such as sensitivity, spatial resolution, and tomographic ripple must be balanced to assess whether the 12PH system, or a redesign of the original 7PH collimator, can yield significantly better diagnosis of coronary disease.

FOOTNOTES

* SIASA, Denver, CO.
† Digital Equipment Corp, Marlboro, MA.
‡ Research Systems Inc, Denver CO.

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Appendix: Fractional Pixel Shifting

Fractional pixel shifting is implemented as a convolution of kernel S (below) with the digital image matrix. To shift the image x pixels to the right and y pixels upwards (where $x \leq 1, y \leq 1$), the matrix S has the form

$$\begin{bmatrix} 0 & 0 & 0 \\ x - xy & 1 - x - y + xy & 0 \\ xy & Y - xy & 0 \end{bmatrix}$$

Shifts in other directions can be implemented with convolution kernels having analogous forms by a simple rearrangement of the matrix elements. This operation is equivalent to a linear interpolation of the data. Repetitive applications of the shifting convolution will result in data that are degraded by smoothing. For this reason, views were not shifted by successive applications of the convolution kernel. Rather, to obtain each tomographic plane, the original views were shifted by integral amounts of simple translation of the image matrices, fractionally shifted using the convolution operation, then combined. By neglecting rount-off errors and elimination of the most peripheral pixels in the image, presumably with insignificant values, the operation preserves the integral counts in the image.

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