SPECT: A Practical Guide for Users

Kim L. Greer, R. Edward Coleman, and Ronald J. Jaszczak

Duke University Medical Center, Durham, North Carolina

The recent surge of single photon emission computed tomography (SPECT) cameras and systems in the commercial market has made unfamiliar equipment available. Current literature tends to present SPECT in terms of mathematics for physicists or clinical results for physicians with little guidance on operating a SPECT system. Yet even the best system is quite unforgiving of sloppy operation. It is not only necessary to have stable, reliable instruments and software but also, the nuclear medicine technologist must be able to perform proper calibration and operation. We review the basic acquisition parameters and data processing techniques necessary for high quality results from a SPECT system.

Compared to conventional or planar imaging, SPECT requires additional effort from the technologist to produce high quality images. Quality control is essential to achieve high resolution in reconstructed images and to prevent the introduction of artifacts. Particular attention to small details is necessary to ensure optimal studies. Artifacts will prevent detection of true defects and can at the same time appear as defects, or structures, where none actually exists. A new routine of calibrations, quality assurance, patient positioning, and data processing (*1*) is necessary to perform SPECT imaging. We present a generalized operating sequence for these unfamiliar with SPECT imaging and suggest methods to avoid or correct some of the more common pitfalls. The literature already notes the necessity of flood (*2*) and attenuation compensation (*3,4*). The technologist must recognize that "short cuts" will systematically degrade the reconstructed images to the point that at best, the SPECT study may be worthless, and at worst, may actually cause harm if it results in misinterpretation by the physician.

Calibrations

For SPECT imaging the camera and the computer must have a common reference, known as the axis of rotation (AOR). Located at the exact center of the camera, this line (Fig. 1) is at a distance equal to one-half the diameter of rotation. At any angle the camera "sees" a line (or point) source mounted on the AOR in exactly the same place, i.e., the center of the image, measured perpendicular to the AOR. The process of adjusting the camera-computer positioning signals to achieve this is known as centering calibration.

Failure to perform centering calibration before acquiring SPECT projection data will lead to (1) loss of resolution, (2) increase in noise and (3) production of circular artifacts in the reconstruction (*5*). With SPECT, centering calibration is one of the most critical quality control procedures to perform. It should be performed and verified before any data acquisitions.

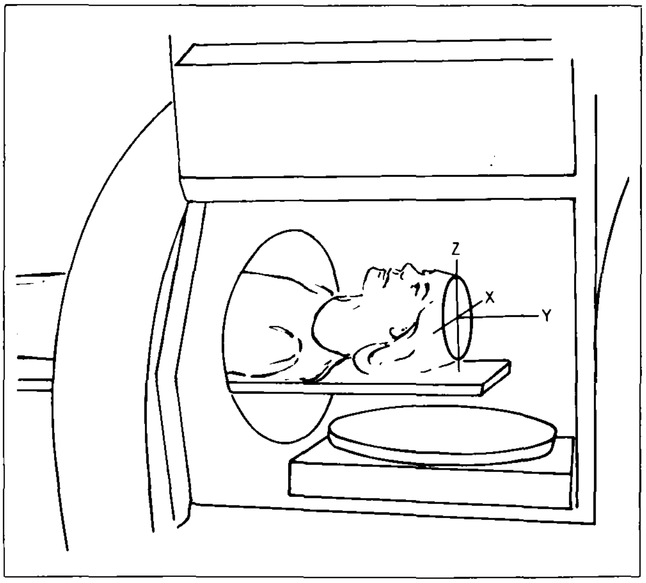

FIG. 1. Diagram demonstrates the X,Y,Z coordinates.

One method for calibrating is to use a point source mounted over the exact physical center of the camera in the "X" direction (perpendicular to the AOR) while making adjustments to the X-offset voltage on the analog-digital converter (ADC). Signal adjustments may also be made with a separate position signal amplifier, either manually or electronically. If miscentered by one pixel, for instance in channel 63.5 rather than 64.5 in a 128-channel wide matrix, the computer image is in error by one pixel, or channel, to the "left" of where it should ideally be. When the camera rotates 180° to the opposite side of the patient, the error is one pixel to the "right" of center. If one channel corresponds to 3.2 mm, a one pixel error in centering results in a total error of 6.4 mm.

Figure 2(A–C) demonstrates a series of SPECT reconstructions of line sources parallel to each other and parallel to the AOR. The images were reconstructed with errors in centering of 0, 1, and 2 pixels. The images show that it would be impossible to determine if Fig. 2(C) actually consists of

For reprints contact: Kim L. Greer, Box 3949, Div. Nuclear Medicine, Duke University Medical Center, Durham, NC 27710.

doughnut-shaped radioactive objects or six point sources made circular because they were improperly acquired.

The clinical significance of this artifact is seen in Fig. 2(D–F), a brain scan. The projection data were acquired with proper centering in this study. The projections, stored on tape, were read back in and shifted 0, 1, and 2 pixels before reconstruction, which is identical to performing three separate acquisitions with various degrees of centering. The lesion in Fig. 2(D) blends into the scalp activity in Fig. 2(F). We allow no more than 0.5 pixel error (1.5 mm) in centering, though typically it is 0.1 to 0.2 pixels (0.3 to 0.6 mm).

A second method for centering calibration involves the use of software to compute the AOR and "shift" the projections in acquisition or reconstruction. This shift or offset could be determined by scanning a line or point source for 360° using an apparatus supplied by the manufacturer. Reconstructions are evaluated to compute the centroid of the source and thus the location of the AOR. Once the centering calibration has been performed and verified, the X-gain, or image size, may be adjusted.

Centering and gain may be adjusted automatically by an electronic circuit that monitors signals from a set of sources mounted at precise locations over the cameras (Fig. 3). The sources are used in both the centering and gain calibrations because their physical location and spacing are known. The X-gain will determine the mm/pixel relationship within a given transverse SPECT image. In some systems the X-gain may be set through imaging point sources with a known distance of separation, varying the X-gain on the ADC or position signal amplifier. The optimal size for most applications is just enough

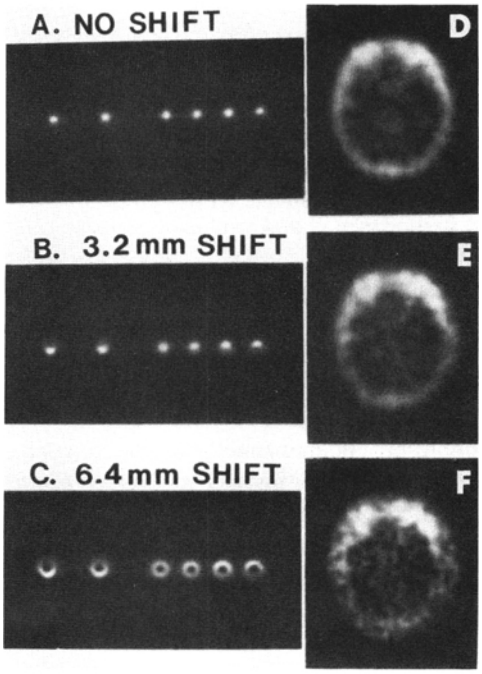

FIG. 2. (A–C) Reconstructed transverse images of seven line sources mounted parallel to the axis of rotation. Centering errors of 0, 1, and 2 pixels (0, 3.2, and 6.4 mm) are represented by (A), (B), and (C), respectively. The brain scans (D–F) demonstrate the effect of reconstructing the same data after shifting it to simulate centering errors of 0, 1, and 2 pixels. The small lesion barely distinguishable from the scalp in (D) blends into the scalp in (E) and (F).

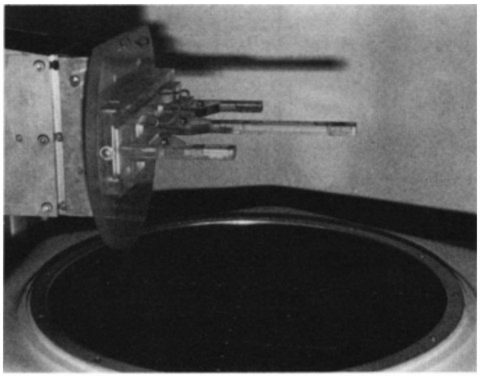

FIG. 3. Centering and X-gain calibration. Three prongs extending over the detector contain a cobalt-57 point source. The long central prong is precisely located over the detector's physical center. The apparatus is dismounted from the gantry when calibration is complete.

gain to fill the computer image matrix with a flood source. The pixel size is important to know in order that attenuation compensation algorithms may correctly calculate the proper coefficients.

Centering and gains are less critical in the Y axis than the X axis. Centering and gains primarily affect the slice level where various organ structures will appear and the slice thickness, respectively. Matched X- and Y-gains will allow for accurate longitudinal (coronal or sagittal) image reconstruction. Matching the X- and Y-centering and gain is critical for two-camera systems.

Failure to perform a standardized quality control exercise will almost certainly result in disappointing and confusing images. The degree of tolerable error depends on many factors including acquisition matrix size, reconstruction matrix size, degree of filtering in the reconstruction, and intrinsic and systemic camera resolution. Several factors that may alter calibration after appropriate centering include mechanically unstable camera supports, interaction of the earth's magnetic field with the photomultiplier tubes (PMTs), and gravitational effects such as the PMTs shifting position on the optical coupling as the camera rotates.

Flood Correction

Correcting for nonuniform field sensitivity is a subject of considerable controversy at the present time. Studies (*2,4*) have demonstrated that flood correction is essential to artifact suppression but there is disagreement on the appropriate correction method. Flood correction of projection data is needed because no camera exhibits perfectly uniform sensitivity over the useful field of view. Cameras equipped with circuits to correct spatial distortions (*6*) eliminate the principal cause of camera nonuniformity. Still, these circuits correct only the intrinsic nonuniformity; they cannot affect imperfectly made or damaged collimator characteristics. Rogers et al. (*2*) and others (*4*) have shown that even small regional variations produce significant, visible circular artifacts in reconstructed images. The artifacts may not always be visible owing to lack of statistics in SPECT slices and to data processing techniques that hide both artifacts and organ structures. Processing techniques that can mask artifacts include coarse acquisition and reconstruction matrices, excessive filtering, and extremely thick tomographic slices.

A quick test to evaluate the need for flood correction is to scan a water-filled cylindrical phantom containing uniform concentration of radioactivity. The acquisition should be performed with the finest matrix size the system provides, the greatest number of angles possible, and the thinnest possible slice thickness. Approximately 5 million counts per SPECT slice are sufficient to demonstrate "hot" or "cold" circular artifacts in a 22-cm diameter cylinder, when reconstructed with the finest matrix size, attenuation correction, and as little filtering as possible. The technologist should verify that each slice appears as a circular disk and that they are uniform in activity throughout. The appearance of "rings" of increased or decreased activity in a slice indicates that flood correction is needed. Again, there is no agreement on the best method of flood correction but the manufacturer or computer software support group should be able to supply a suitable algorithm.

To perform a statistically valid flood correction, the flood correction images must contain a high number of counts. A separate image is required for each detector, and for each collimator, when changed. The floods, acquired only after the centering calibration, may contain 40–50 million counts each for a 128 × 128 matrix. It must be remembered that 40–50 million counts spread over a circular field in a 128 × 128 matrix results in only 1200–1600 counts/pixel. The number of counts necessary for correction will depend on the area of usable field of view, matrix size, algorithm, and degree of precision required. Figure 4 shows the effect of various count flood images on flood corrections.

The length of time required to acquire the flood images is usually 20–25 min. A well-mixed flood phantom filled daily with 6–7 mCi of Tc-99m (LEAP collimator) is sufficient for the desired number of counts in a reasonable time. This allows approximately 30 cps/cm^2 to be detected. Correction images taken with one energy analyzer setting can be used for studies

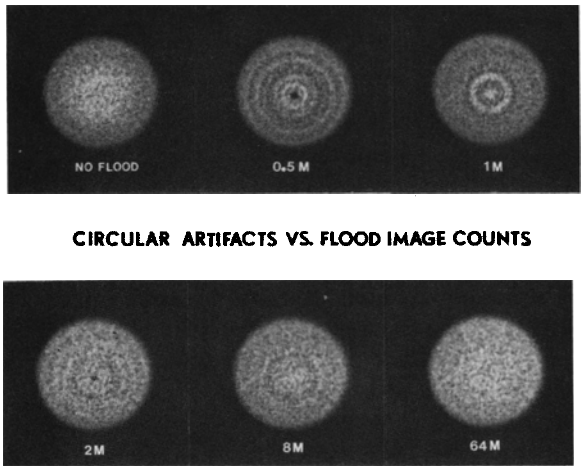

FIG. 4. Reconstructions of uniform activity cylinder at the same slice level. Same projection data were corrected from flood images that contained varying numbers of counts. The top left image was not flood corrected. The flood images used contained 0.5, 1, 2, 8, and 64 million counts. The increased counts per pixel with the 64 million count image allows the statistical certainty to perform this type of compensation accurately.

acquired at a different energy setting, for example, Tl-201 or I-123. Every camera manufacturer performs corrections for spatial distortion, uniformity correction, and pulse height analysis/energy discrimination by some different method. Thus, we recommend a test scan of a uniformly distributed, water-filled cylinder containing a radionuclide, for example Ga-67 or I-123, using flood images from Tc-99m.

Scattering medium between the sheet source and detector is not used. Phantom reconstructions are free from circular artifacts when flood images that do not include scatter are used. The amount of scatter to include and the method and material for Compton scatter generation have not been validated.

Another source of nonuniformity relates to the collimator. Approximately 10 million counts in a 128 × 128 flood image is sufficient to visualize collimator imperfections. These may be enhanced by computer controls, specifically "windowing" techniques that vary the displayed counts/pixel brightness and number of gray tones. Statistical variation is too great in a 1–2 million count flood image to visualize small but significant changes in regional sensitivity. In general, we believe any visible nonuniformities are more than sufficient to generate serious artifacts and should be compensated for. Asymmetric energy analyzer settings, damaged collimators, and poor optical coupling can all contribute to lack of uniform field sensitivity.

Two assumptions made regarding flood correction are (1) camera response is independent of camera angle (gravitational, magnetic effects) and (2) camera response is independent of count rate. These assumptions should be validated upon installation. From a practical standpoint, little can presently be done to compensate for these effects even when known.

Initial Set-Up

Once a SPECT system is installed, a rigorous series of calibrations and phantom tests should precede patient studies. The first step is to perform the usual acceptance tests for a new camera used in the conventional manner. Field uniformity and linearity, full width-half maximum (FWHM), full width-tenth maximum (FWTM), and bar phantoms are some of these tests.

Before any SPECT phantom tests, the technologist must perform centering and gain calibrations for all detectors and acquire the flood image(s) used in the reconstruction program to correct for nonuniform camera sensitivity.

A good test for the initial evaluation is a SPECT scan of several small (< 1 mm i.d.) glass capillary tubes filled Tc-99m sodium pertechnetate, mounted first in air and then in a water-filled cylindrical phantom. These tubes are positioned parallel to the AOR with a separation of 5 to 10 cm. One of the capillary tubes should be mounted near the AOR. The photopeak is set according to the manufacturer's instructions. The same collimator that will be used in the clinical studies should be used for this. A standard radius of rotation is necessary for the goal of quality assurance. Begin the SPECT scan sequence, again following the manufacturer's instructions, using the finest acquisition parameters available (i.e., most number of angles, matrix size, etc.), and the parameters the manufacturer recommends. Adjust the scan length, or rotation time, to collect at least 75,000 counts in each line/slice. Reconstruct the scans with the finest recommended parameters. On visual inspection, each line in the reconstructed images should appear as a point, or cross-section, of the line. If they appear as rings, recalibrate the system and repeat the scans. Analyzing each point for FWHM will yield an approximate value for system resolution in the SPECT mode. Also, the lines mounted in water in a 20-cm cylinder should be scanned.

Next, scan a cylindrical, water-filled phantom containing approximately 1 μCi/ml. A 20–25-cm diameter cylinder, whose length extends over several slices, is sufficient. Scan the cylinder in a manner similar to the line sources experiment. Also, collect the data for attenuation correction. These data are used to generate a body counter, or outline, which is critical (*7*) to correcting for photon attenuation. Expect a relatively flat count profile through the cylinder. A profile that smoothly decreases in activity from the periphery to the center of the cylinder is likely caused by lack of attenuation correction, use of a too small value for the attenuation coefficient, or differences in the thickness of the flood source. If the attenuation coefficient is too great, the activity near the center will be greater than at the edge of the cylinder.

To estimate a system's ability to detect lesions, scan the cylinder again, under the same conditions, except this time mount nonradioactive spheres inside the cylinder. Include sphere diameters of 1–4 cm (Fig. 5); these will simulate cold lesions in a liver scan.

These few experiments will help to assure that patient data are as reliable as possible. Artifacts must be recognized and corrected. Daily, weekly, and quarterly routines of quality control will alert the SPECT operator to problems.

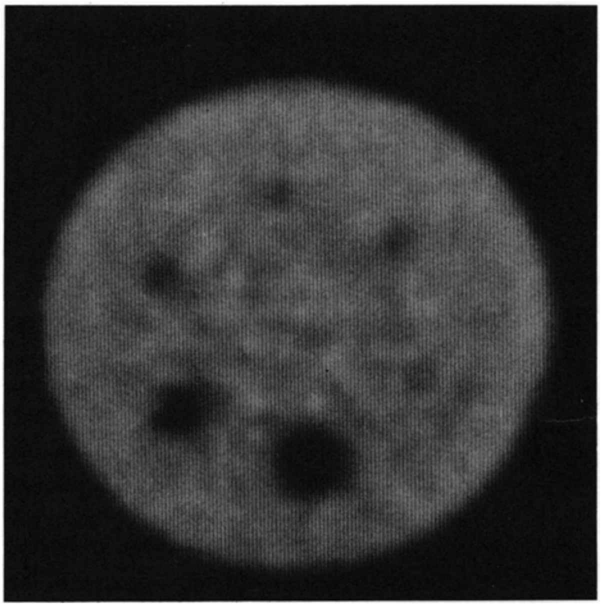

FIG. 5. Reconstruction of a phantom containing six nonemitting spheres in uniform activity background. This type of phantom can provide much useful information concerning artifact presence and ultimate ability to visualize a given size "lesion" with a given count density to simulate clinical studies.

Clinical Studies

Clinical SPECT examinations require more precision and attention to detail in patient positioning and data acquisition than planar imaging.

Patient positioning is more critical in SPECT for several reasons. The patient must be positioned in the X, Y, and Z dimensions (Fig.1) relative to the camera. The camera-to-patient distance should be as close as possible but the radius of rotation must be large enough so that the camera will not touch the patient as it rotates. The camera orientation should always remain parallel to the axis of the rotation (AOR), i.e., without any tilt of the detector surface toward the patient's head or feet.

The area of interest can be positioned in the Y direction by positioning the imaging table under the camera. The portion of the detector used must ensure that the entire organ is imaged. Some systems use only the central half of the camera in the Y direction (parallel to the AOR) and usually all of the detector in the X direction (perpendicular to the AOR).

Once this positioning is satisfactory, the camera radius of rotation can be adjusted to minimize the patient-to-detector distance. Since human cross-sections are generally elliptical, the detector distance will vary as the camera rotates around the patient. Finding the minimum possible distance can be accomplished by moving the camera in or out and elevating or lowering the support pallet as needed. Rotating the camera manually once or twice will aid in determining if further adjustments are necessary before actual data collection. "Step and shoot" systems allow the camera to be moved in or out from the AOR during the scan. Extreme care must be exercised with this practice, because with such a counter-balanced system the camera will also change position relative to the Y axis of the patient.

Patient comfort is essential to reduce patient motion once scanning begins. Motion affects SPECT scans in several ways. As in planar imaging, motion during a SPECT scan causes a general reduction in resolution and contrast and may also result in streak artifacts.

If possible, the patient should fold his arms overhead, supported with sponges to reduce fatigue, when thoracic and abdominal exams are being performed. Slight degradation has been noted in scans of patients whose arms were down. Having the arms down will increase the detector radius of rotation, reduce resolution, increase photon attenuation, and even possibly distort the body contour data used for attenuation compensation. It is important to remember that SPECT images are produced from numerous planar images and that, generally, anything that affects the planar (projection) data will affect the SPECT images.

For head studies, the patient should be positioned under the camera so that the entire area of interest, and nothing else, is visible to the camera at all angles of camera view. Shoulders can interfere with achieving a minimum radius of rotation. This situation can be helped by telling the patient to relax his shoulders while he stretches his arms towards his feet. As with any study, the entire organ, or at least the slices of interest, must be completely visible from any angle.

Data Processing

Technologist interaction is required in three general areas of SPECT imaging: data acquisition; image reconstruction; and image display. Better images will generally result from longer acquisition times, finer image matrix size, and a larger number of angular samples.

The image matrix size is not necessarily the same for SPECT and planar images. A SPECT acquisition matrix may consist of 128 samples per projection, i.e., 128 pixels wide, to minimize aliasing (sampling) artifacts. Acquisition time is based on the camera rotation time through 360° which is variable from 2 to 26 min. Gantry motion may be either continuous or "step and shoot," depending on the type of equipment available. Body contour, for attenuation correction, may be acquired from scatter information, using an energy analyzer window setting of 33% (FWHM) centered at 106 keV.

The filters usually used for brain or liver scans have cut-off frequencies that correspond to approximately 1.6 or 1.1 cycles cm^{-1}. The attenuation coefficient used in brain and abdomen scans has a value of 0.12 cm^{-1}. A value of 0.09 cm^{-1} is used for lung scans. Carefully choose the proper filter and attenuation compensation.

Total acquisition times for SPECT scans compare favorably with planar studies. The patient is positioned just once for the SPECT study but many times for a planar study. Also, any view possible with the conventional camera can be approximated with SPECT images.

Summary

Training programs in nuclear medicine technology will soon need to include education in SPECT applications. The potential for errors in SPECT imaging is as great as the potential benefits. We re-emphasize the need to perform phantom studies before clinical use and to determine appropriate scan/reconstruction protocols for each individual system and application. SPECT technical operation requires greater attention to detail than planar imaging does.

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