Single-Photon Emission Computed Tomography in the Year 2001: Instrumentation and Quality Control

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Objective: SPECT instrumentation is more complex than that used for whole-body and planar imaging, and requires careful quality control to ensure optimum performance. Conventional and new hybrid SPECT imaging systems (coincidence and SPECT/CT) will be discussed. New imaging detector materials such as LSO and CZT will also be discussed, along with their potential advantages. Finally, basic SPECT quality control will be reviewed. After reading this article, the nuclear medicine technologist should be able to: (a) explain the use of single and multihead gamma cameras for SPECT imaging; (b) have an understanding of the potential of new hybrid SPECT imaging systems; (c) be aware of future developments in SPECT imaging technology; (d) understand the requirements for SPECT quality control, including field uniformity and center of rotation corrections; and (e) explain the benefits of using phantoms to augment SPECT quality control.

Key Words: SPECT; CT; instrumentation; quality control

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SPECT INSTRUMENTATION

Although several prototype devices for SPECT have been developed over the years, modern commercially available SPECT instrumentation is based on the rotating gamma camera, either single or multihead. For SPECT systems, the trade-off in SPECT acquisition for the nuclear technologist is between spatial resolution and sensitivity. Better spatial resolution is desirable to visualize detail, so a high-resolution collimator would seem appropriate; however, the statistical noise content of the SPECT study may be high. Better sensitivity is desirable, to reduce the noise in the images, which would suggest the use of a higher sensitivity (with concomitant lower spatial resolution) collimator. To further complicate the choice, there are additional methods to improve the spatial resolution of a scan, which include zoom factors for the projection images (smaller millimeters per pixel), the use of fanbeam or other focusing collimators, and the use of noncircular orbits to minimize the distance from the collimator to the patient (*1,2*). Noncircular orbits (NCO) allow the camera to remain close to the patient's body during a scan, improving patient-to-camera distance, but do require some extra setup time to define the limits of radial motion and could require a more complicated center of rotation (COR) correction for the reconstruction algorithm.

Singlehead SPECT Systems: Camera Considerations

The benefits of a single-head camera SPECT system are that it is relatively inexpensive compared with multihead SPECT systems, and quality control (QC) is fairly straightforward. The disadvantages of single-head SPECT systems include relatively low sensitivity compared with multihead systems and thus, a generally longer patient acquisition time. However, there are many single-head systems in use today performing perfectly adequate SPECT imaging.

With single-head SPECT systems, scan times are seldom less than 15–20 min and are frequently 30 min or more. Typically, acquisition times of more than 30 min cause significant patient discomfort and may actually provide scans inferior to those done using a shorter acquisition-time-scan, due to a higher degree of patient motion. For single-head systems, the use of an all-purpose collimator with an NCO orbit is frequently chosen as the best compromise to keep the scan time under 30 min. However, the nuclear technologist must choose the set of acquisition parameters that will give the "best" information for an acceptable scan time and dose to the patient.

Multihead Camera SPECT Systems

Camera systems with 2 or more heads surround the patient with more detectors and offer more optimal spatial resolution/sensitivity characteristics than are available with

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a single-head system (*3*). These combinations assume that the data from different heads are matched in gain, orientation and offset, so that the data can be combined. Disadvantages of multihead systems are that they are more expensive than single-head cameras because the cost of collimators goes up substantially (now you have collimator "sets"), and the QC procedures must ensure that adding the data from different heads can be performed without artifact. These systems also require more elaborate carts or automated devices to aid in changing the 2 or 3 collimators per system.

Multihead SPECT systems are typically 2-head for cardiac studies or 2- or 3-head for other SPECT applications such as brain, bone, and oncologic SPECT applications (i.e., ^{67}Ga, ^{111}In, or ^{99m}Tc agents). These multihead systems provide shorter SPECT acquisition times over single-head systems. A practical example is given with the following: Let a single-head camera with a low-energy, all-purpose parallel hole collimator acquire N counts in a 30-min scan. With the same type collimators, a dual-head camera can obtain 2N counts in 30 min (twice the sensitivity), or obtain N counts in only 15 min (twice the throughput). This extra sensitivity permits higher resolution collimators to be used and still have an acceptable number of counts in the allotted scan time. If a 2-head system used high-resolution parallel-hole collimators that had 60% of the all-purpose collimator sensitivity, then a 30-min scan with these collimators would have 1.2N counts and higher spatial resolution. Depending on the study type, this scan could contain significantly more information than the N count acquisition with the all-purpose collimator. The use of fanbeam or other focusing collimators would further improve the spatial resolution and the sensitivity, but at the expense of field-of-view (FOV) and complexity. It is fairly well-accepted in the nuclear medicine community that for cardiac studies, dual-head cameras at 90° to each other is the optimal configuration, and allows most studies to be performed in about 15 min with excellent image quality.

The use of a 3-head system provides still more acquisition combinations for the operator to choose from for noncardiac applications. For the same 30-min scan mentioned above, one could get 3N counts using the same collimator types (even higher sensitivity), or N counts in 10 min with even greater throughput. With this increase in sensitivity, there are ample counts to allow the use of higher resolution collimators and improve the diagnostic content of a scan. In fact, the gains of higher resolution and greater sensitivity with focusing collimators for 2-head systems would be even greater for 3-head systems. The use of 3 fanbeam collimators of appropriate focal length would result in better spatial resolution and sensitivity, and still have a FOV large enough for many, but not all, patients.

The benefits of 3-head systems encouraged the examination of 4-head systems. However, the expense of the camera system goes up substantially with each head, so the improvement in performance must be significant to merit the additional cost of detectors and sets of collimators. Also, with 4 heads, it is harder to use the device as a general-purpose imaging system. Four large detectors would have to have a large scan radius to take full advantage of the large FOV (LFOV). For brain imaging, much of the LFOV detector area would be wasted, which would be undesirable for such a system. A commercial system with 4 smaller heads was developed, but had been taken off the market at the time of this writing (*4*).

New Imaging Devices

Historical Review of Non-Traditional Anger Camera SPECT Systems. To avoid the problems with SPECT systems having many moving detectors, SPECT systems in which only the collimator moves have been developed in the academic community.

SPRINT II (*5*) was a single slice system designed for brain imaging that has approximately 100 NaI detectors on a fixed ring. The collimator is a rotating aperture ring containing 8 slits. A resolution of 8 mm has been obtained, but only 1 slice at a time can be imaged. Another unique design is a hollow cylindrical NaI crystal detector that uses a rotating collimator inside the crystal (MUMPI III) (*6*). Photo-multiplier tubes (PMTs) are attached to the crystal exterior. The crystal and collimator surround the brain. The advantages are that 4 slices are acquired simultaneously, and that the high sensitivity allows data to be acquired quickly. Although one disadvantage of this design had been the cost and difficulty of obtaining a cylindrical NaI crystal, curved NaI crystals are more readily available today (e.g., CurvePlate™; Bicron Corp., Newbury, OH). A similar annular crystal detector, ASPECT 3000 (Waltham, MA), has been developed by Digital Scintigraphics for brain imaging. This is a 3-ring system, with 21 PMTs per ring. ASPECT has an annular collimator system consisting of 3 equal-sized, parallel hole collimators rotating incrementally. Modular gamma cameras are also being investigated (*7*). This type of camera is composed of modules having 4 PMTs per NaI crystal. Several of these modules would be positioned around the patient. This design would have lower hardware costs, as well as higher counting rates than a typical gamma camera. However, edge packing and uniformity would be significantly greater problems in this design. A maximum likelihood algorithm is being used for reconstruction. It is not known when this approach will generate images of an extended object comparable to those of current gamma cameras.

Current Progress in Hybrid SPECT Systems. Hybrid SPECT systems, including coincidence cameras (which can perform both SPECT and PET imaging) and hybrids that incorporate a CT scanner along with a SPECT system, have received much attention. Several manufacturers are developing combined SPECT/CT systems that hold promise for improved attenuation correction. In addition, these systems allow a CT scan to be performed along with the SPECT scan for image registration or fusion of the SPECT scan with a high anatomic resolution CT scan, albeit at a higher device cost than a SPECT system alone. Adequate attenuation maps (μ-maps) with existing moving or stationary transmission sources are not fully optimal. The CT high-photon flux may provide superior attenuation correction for cardiac SPECT. Moreover, with receptor and monoclonal antibody imaging in oncology (Prostascint™, antiCEA imaging, for example), the superb spatial localization provided by the CT scan may aid in determining the location and extent of disease identified by valuable, but lower resolution, SPECT studies.

A coincidence imaging system is a dual-head gamma camera system in a 180° opposed configuration. These systems can acquire a conventional single-photon SPECT scan, as well as coincidence 511-keV events from a positron-emitting agent for PET imaging. The relatively low stopping power of NaI at 511 keV, compared with crystals with higher stopping power such as BGO, requires the use of thicker NaI crystals (½–⅝-in or more). The dilemma in using coincidence cameras is that the thick NaI crystals required to stop adequate numbers of high-energy 511-keV photons degrade single-photon performance at ^{99m}Tc and ^{201}Tl energies. In essence, the thicker crystal requires the low-energy light photons to traverse greater distances from the location of the gamma ray interaction in the crystal, resulting in a more diffuse signal reaching the PMT. Thus, a greater amount of statistical noise is introduced at a greater moment arm from the event, with a concomitantly greater error in positioning the event in the camera. The Anger principle, by which events are positioned in the scintillation camera, is analogous to a center of mass calculation. The event is positioned by the product of the amount of light seen by a PMT, times the moment arm, determined from the PMT weighting coefficients, and divided by the total amount of light collected for the event. If light must travel larger distances through the crystal, a noisier signal is seen at each PMT, which increases the uncertainty in positioning the event and results in poorer intrinsic spatial resolution.

One recent development is the Starbrite™ NaI crystal from Bicron Corporation (Newbury, OH). The Starbrite™ crystal is 1-in thick, providing improved stopping power for 511-keV photons, but is composed of 2 NaI "layers," ½-in of uniform crystal with the second ½-in slotted to improve image spatial resolution at low, ^{99m}Tc, energies. The slotted NaI crystal portion of the Starbrite™ improves the light response function of the crystal to make it appear that the light photon events are incident in an apparently "thinner" camera crystal. In other words, the slots bring the light from the scintillation event into the PMTs closer to the position where the event occurred in the crystal. Recent evaluation of one such system with the Starbrite™ crystal produced ^{99m}Tc and ^{201}Tl SPECT scans, which were only slightly degraded from those acquired by a ⅜-in dual-head gamma camera system by direct comparison of the same patient on both systems *(S.M. Spies, MD, Oral Communications, Sept 2000).* An example of an initial bone scan using the Starbrite™ crystal system (Duet™; Siemens Medical Systems, Hoffman Estates, IL) is shown in Figure 1, with the same patient acquired on a conventional ⅜-in gamma camera (e.cam™; Siemens Medical Systems) for comparison. These thick, slit crystals have the potential to perform high-energy imaging with little degradation of low-energy performance.

The use of a new scintillation material, lutetium oxyorthosilicate (LSO), is being investigated. LSO has a much higher stopping power than NaI for 511-keV annihilation photons and nearly 75% the light output of NaI, but significantly more light output than BGO (*8*). Moreover, it is a faster scintillator, thus allowing higher counting rates than both NaI and BGO crystals. Table 1 provides the relative characteristics of nuclear medicine scintillation crystals. LSO is naturally radioactive; if used for single-photon work, corrections must be applied. Nonetheless, its use in PET and PET/SPECT imaging systems has been realized, and prototype units are undergoing initial clinical trials *(R. Nutt, PhD, Oral Communications, Nov 2000).* Currently,

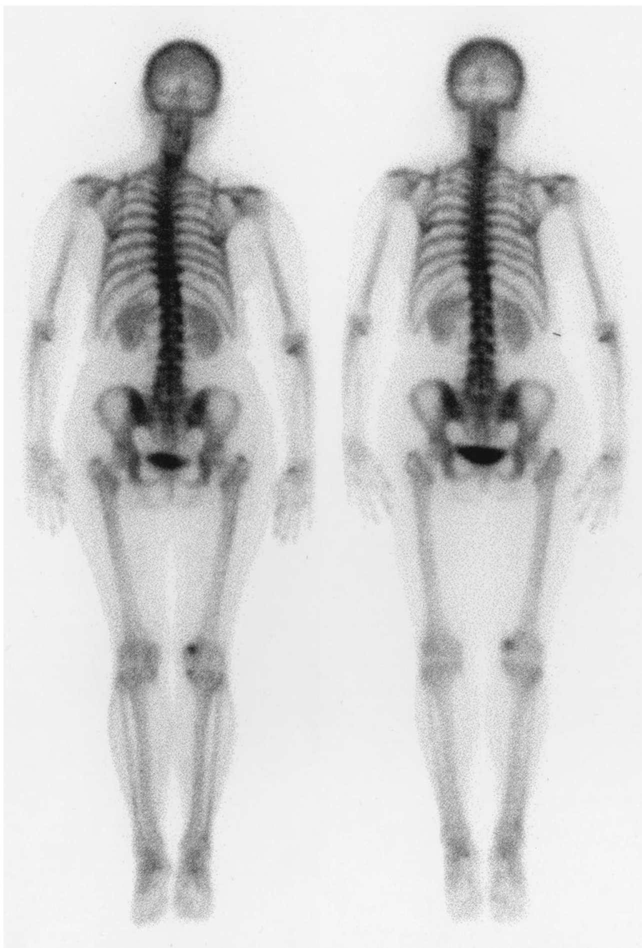

FIGURE 1. Posterior whole-body bone scan (right) obtained using a thick, slit crystal gamma camera system (Duet™; Siemens Medical Systems), compared with the same scan (left) acquired on a conventional ⅜-in gamma camera system (e.cam™; Siemens Medical Systems).

TABLE 1
Characteristics of Nuclear Medicine Scintillation Crystals

Crystal time	Effective Z #	Light output (relative to NaI)	Scintillation decay
NaI	50	1.0	230 nsec
BGO	72	0.12	300 nsec
LSO	65	0.75	40 nsec

there are production facilities for LSO in Knoxville, TN and in Florida, and the supply of LSO should be significant within the next year *(R. Nutt, PhD, Oral Communications, Oct 2000)*. Hybrid systems of LSO may prove optimum for both high-energy (511-keV coincidence) imaging applications and conventional ^{99m}Tc SPECT applications.

Stopping power for medium- and high-energy imaging is important. However, if very high spatial resolution is desirable for ^{99m}Tc agents, a new version of the solid-state detector cadmium telluride, CZT (cadmium zinc telluride), has shown considerable promise as a new nuclear medicine gamma camera detector material. CZT has desirable energy resolution characteristics and can be used at room temperature. This is in contrast to most solid-state detectors, which must be cooled by liquid nitrogen to reduce detector noise. When CZT is made into mosaics of small detectors, outstanding spatial resolution may be obtained. These devices can be made small and modular, a further advantage for nuclear medical imaging.

Over the years, research efforts to overcome count sensitivity limitations have led to experimental devices such as the Compton scatter camera, which is collimator-less and uses Compton angles to position gamma ray events in a conventional camera with the aid of a high-resolution, solid-state detector. Gas-filled multiwire camera prototypes have been built for very high count rate imaging applications, such as first-pass imaging.

In all, the potential for new imaging devices in the next 5 y is great, and is categorized in outline form:

1. New "Small" gamma cameras.
   a. CZT and other solid-state cameras.
   b. CsI or NaI crystals with photodiodes or position-sensitive PMTs.
2. New "Large" detector SPECT systems.
   a. NaI crystals.
      i.) Curved NaI (Polyscint™; CurvePlate™; Bicron Corp.).
      ii.) Slit crystals (e.g., StarBrite™; Bicron Corp.).
   b. Gantry mounted cameras.
   c. LSO
   d. Composite mosaics of CZT modules.
3. SPECT/CT Hybrids.
4. Coincidence cameras (511-keV) and SPECT/PET Hybrids.
   a. NaI thick crystal $\geq$ 1 in.
   b. NaI slit crystals (StarBrite™; Bicron Corp.).
   c. LSO systems.
   d. GSO systems.
5. Experimental devices.
   a. Compton scatter camera.
   b. Multiwire gas-filled proportional cameras.

QUALITY CONTROL FOR SPECT

Quality control (QC) performed on nuclear medicine cameras provides the confidence to technologists and physicians that a SPECT scan supplies an accurate representation of the radioisotope distribution in the patient. Whereas planar detector quality control measures help ensure high-quality planar imaging, SPECT imaging requirements place more stringent and additional performance requirements on a detector system (*11–20*). In fact, it is possible that a camera that is functioning well as a planar instrument may produce artifacts in the SPECT image, which may adversely affect patient management (*16*).

Guidelines for the frequency of QC testing and for the choice of tests to be performed at certain intervals have been given in the literature, and are best reflected in the NEMA standards (*10–11*). The SPECT corrections that have the most significant effects on reconstructed image quality are uniformity correction and COR correction. Other tests include pixel size (gain) calibration, linearity measurements, rotational sensitivity, mechanical alignment, energy resolution, and counting rate capabilities.

Uniformity Correction

One of the most common and most severe reconstruction artifacts is the concentric ring or "bull's eye" artifacts caused by regional sensitivity variations in the projection images (*18, 19*). These variations are caused by camera spatial nonlinearities, differences in crystal thickness or energy response, and collimator septum disparities (*14*). Flood images taken with the collimator in place are called extrinsic flood images, or extrinsic floods. Flood images taken without the collimator in place are called intrinsic floods and show the state of the detector tune, linearity correction, and energy analyzers. SPECT uniformity correction adjusts for the nonideal collimated detector, so extrinsic floods are used. Some manufacturers shortcut this process by having a "collimator map" and rely on intrinsic flood correction to provide the 1% uniformity requirement (in conjunction with the collimator map). This method will work if the collimator is not damaged in any way after the collimator map is obtained.

The frequency of uniformity correction calibrations seems to be changing, perhaps reflecting the more developed and stable electronics used in modern gamma cameras. Earlier works (*13*) recommend daily (extrinsic) flood correction. Graham (*12*) recommends weekly corrections. Halama and Madsen (*14*) recommend daily intrinsic floods and weekly extrinsic floods. Careful monitoring of the changes in the acquired floods will be more useful in determining the appropriate interval for uniformity correction for specific SPECT systems. The interval between subsequent acquisitions of the uniformity correction may be a week or 2, and could be as long as a month for the optimum correction if the stability of the system has been evaluated for the longer time interval.

A circular uniformity artifact in a reconstructed slice is generated from a pixel or group of pixels having increased or decreased sensitivity. The size of the artifact depends on the image statistics, as well as on the diameter of the source distribution, and inversely on the square root of the distance of the sensitivity variation from the COR. This amplification of sensitivity variations places severe requirements on the uniformity correction to reduce the generation of circular artifacts at or below the magnitude of the noise effects in the reconstructed image. It is recommended that flood images taken for uniformity correction have random fluctuations and corrected variations in camera response with rotation kept below 1% (*18*). Current camera useful FOV (UFOV) nonuniformities are approximately 3%, which requires a "correction" to reduce the nonuniformities to under 1%. This suggests that at least a 30 *M* count flood for 64 × 64 images or a 120 *M* count flood for correction of 128 × 128 images (about 10,000 cts/pixel in the flood image).

Two types of flood sources can be used for uniformity correction (*19*). ^{57}Co flood sources have the advantages of being solid (no spills), light, and relatively long-lived, with a gamma ray energy close to ^{99m}Tc, and can be commercially purchased. It is imperative that the source be checked to ensure that it is truly uniform to 1% (some vendors claim 1%, but do not always meet this specification—care should be taken and the sheet source evaluated after purchase). Liquid flood sources have also been used and have the advantage of being filled with different isotopes; ^{99m}Tc and ^{201}Tl, for example. Moreover, liquid-filled uniformity sources are much less expensive and do not require replacement every year. However, care must be taken to uniformly mix the isotope, and to fill the source with a precise amount of water in such a way as to prevent bulging or collapsing the center of the flood, and to avoid bubbles that would act as defects in the flood. Some flood sources have a small raised chamber out of the UFOV to collect air bubbles and keep them from generating potential artifacts. Note that with some cameras, uniformity calibration for higher energy sources, (e.g., ^{67}Ga, ^{131}I) may not be able to be performed using ^{99m}Tc or ^{57}Co energies. Calibration may be required using the higher energy sources. The manufacturer's specifications for medium and higher energy imaging should be reviewed.

The camera should be evaluated for a rotational dependence of the flood images (*11,20*), which would indicate a systematic change in sensitivity/uniformity with gantry angle, perhaps due to magnetic fields near the camera. This check would ensure that an accurate uniformity correction could be performed with only 1 correction matrix, and that different gantry angular ranges would not require their own correction matrix.

Another method of reducing uniformity artifacts is to use noncircular orbits (*21*). Use of noncircular orbits can reduce the amplitude of the central ring artifact by factors of up to 20, as well as give increased spatial resolution in the reconstructions, by keeping the collimator closer to the patient than a circular orbit would permit. Noncircular orbits (NCOs) include several motions, including elliptical rotation and circular rotation with translation. Some NCO systems sold today move both the heads and the patient bed during the NCO movement. However, with some systems, the NCO motion is accomplished by only radial in-and-out motion of the camera head, thus ring artifacts may still be present. Multiple head systems also reduce ring artifacts due to the fact that 2 or more heads are involved, making it less likely that nonuniformities will occur in the same places.

COR Correction

To accurately reconstruct projection data, the reconstruction algorithm must know the relation between the physical or mechanical axis-of-rotation and the center of the projection images. The correction that relates the axis-of-rotation's location to the center of the projection images is called the COR correction. Without this correction, the projection data are improperly positioned when back-projected into the reconstructed slices, leading to a loss of spatial resolution at best, or artifacts at worst. If a point source is reconstructed with an accurate COR correction, it appears as a single point in the reconstructed slice with resolution appropriate for the collimator used, distance of the point source from the collimator face, attenuation of material surrounding the source—if any—and the energy of the radioisotope used. If the COR is inaccurate, then the resolution of the reconstructed source will deteriorate (get larger) until the error is so large that the source is reconstructed as a donut (*18,22*). If present in the SPECT system, the large, uncorrected COR errors in clinically acquired studies cannot only throw away a significant amount of otherwise attainable spatial resolution, but also generate several artifacts.

To begin COR corrections, the camera face must be parallel to the axis-of-rotation, or such that the normal to the axis-of-rotation is also normal to the detector surface (i.e., no tilt). This alignment will prevent data provided by several ideally aligned slices in a patient from being smeared together by improper orientation of the detector head. Also, the amount of smearing would vary with rotation about the patient. Lastly, the reconstruction algorithm positions events with the assumption that the camera face is parallel to the axis-of-rotation, so a violation of this assumption cannot properly position data in the reconstruction.

COR corrections are performed by placing a point or line source(s) in the FOV of the camera, and then performing a SPECT scan of the sources. The advantage of a line source is that the COR can be measured for each slice to be reconstructed, so that small pointing or angular imperfections in the collimator could be accounted for if the scan and COR correction were measured at the same scan radius. COR corrections are essential to achieving high-quality reconstructed images from SPECT camera systems. It is recommended that COR corrections be done weekly or biweekly. Some cameras have a bubble level to help indicate when the detector head is properly aligned with the axis-of-rotation. The level can only be used for this purpose if the camera axis-of-rotation and patient imaging table have been horizontally aligned. Once aligned, this method is a quick way to check that the detector head does not have an appreciable axial tilt. Periodic checks of the system alignment may be necessary to see if supports or shims have been compressed, or if a new building or floor may have settled.

Performance Evaluations Using Phantoms

To really evaluate the uniformity and COR corrections, one needs to be able to generate high-quality projections and reconstructions and see how the different corrections and operations of the reconstruction process affect the signal-to-noise ratio (SNR), contrast, and spatial resolution of the high-quality images at different stages in the processing steps (including reconstruction). Performance evaluations using phantoms permit high-resolution studies of camera/system performance with a known object distribution, without the blurring effects of patient voluntary and involuntary motion, and without count limitations caused by patient dose considerations. Another important consideration is that a phantom can be repeated and compared with previous acquisitions, to check the camera performance over time, or after system or software upgrades. The importance of this kind of testing is evident in the number of phantoms now being sold for performance evaluations of tomographic systems. An example of a SPECT phantom image is given in Figure 2. To the left, "cold" rod "pie slices" allow spatial resolution to be assessed, whereas, the "cold" spheres to the right allow both resolution and image contrast to be evaluated. Notice a slight ring artifact in the left image, midway between the center and edge of the phantom; this is likely due to the high count acquisition (i.e., the ring would likely not be visible in a clinical study, at clinical count statistics).

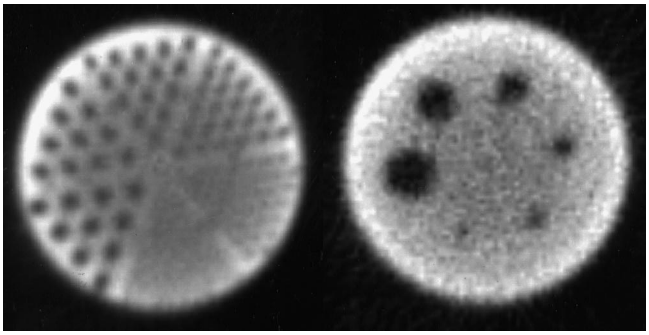

FIGURE 2. SPECT phantom acquisition (Data Spectrum Corp., Hillsborough, NC) with "cold" rod pies shown to the left and "cold" sphere to the right (all 6 visualized). The acquisition parameters were 128 × 128 matrix, 128 views, 1.23 Zoom, 3.9-mm pixel size, approximately 1 million counts per view, 128 million total counts using a dual-head camera (e.cam™; Siemens Medical Systems). This protocol represents a "super" high-resolution/count statistics acquisition.

No evaluation is complete until the entire system acting as a whole has been tested thoroughly. However, for some evaluations, it would be advantageous to selectively test only certain parts of the system. For example, to evaluate a new or modified reconstruction code, it would be useful to use either a simulated (and therefore precisely known) or previously selected projection dataset as input, and to compare quantitatively the output of the reconstruction with the output of the previous algorithm. A check of the COR correction could be performed by simulating a projection dataset with a known offset, inputting the offset into the COR correction algorithm, and evaluating the differences between the COR corrected, shifted dataset, and the original, unshifted dataset. Another check of the COR correction would be to image 1 or more point sources positioned on the axis-of-rotation. In the absence of an attenuating medium, the reconstructed resolution of the point should be approximately equal to the resolution of the point source at a distance equal to the scan radius from the collimator face, using a ramp reconstruction filter. If the reconstructed resolution using a ramp filter is significantly wider or worse than the collimator resolution at the scan radius, then the COR correction should be investigated.

The uniformity correction can be evaluated by imaging a uniform phantom or a phantom with a uniform section. If the source distribution in the phantom is uniform, then the transverse slices through the phantom should also be uniform, but modified for the attenuation of the phantom itself. As mentioned above, the effects of the detector nonuniformity are circular ring artifacts. The visibility of these artifacts will depend on the relative noise amplitude of the slice data, compared with the amplitude of the uniformity fluctuations of the detector. If the slice data have noise fluctuations of 30%, and the uniformity fluctuations of the detector are of the order of 2%, then the uniformity fluctuations will probably not be noticed. However, if the slice data have fluctuations on the order of 1%, and the uniform fluctuations are about 3%, then the circular artifacts will be seen. If a uniformity correction is performed on the projection data before back-projection, and these reconstructed data are now compared with the original uncorrected reconstructions, then the size of the ring artifacts should be reduced. Due to the statistical nature of the flood data, however, the noise in the corrected image will have increased somewhat.

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