The Influence of Collimators on SPECT Center of Rotation Measurements: Artifact Generation and Acceptance Testing

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Misalignment between the electronic and mechanical axes of rotation will result in artifact generation and image degradation during single photon emission computed tomography (SPECT) reconstruction. Acceptance and quality control testing procedures have not emphasized the variability in center of rotation (COR) measurements caused by collimators and the need to verify uniformity across the full collimator field of view (FOV). Variation from the mean COR across the FOV was tested in four different collimators using multiple point source acquisitions. The mean COR was different for each collimator and two of the four had a >0.5 pixel difference from the mean COR on some area of the FOV. This variation makes these collimators unacceptable for SPECT acquisition. Thus, initial acceptance testing of SPECT collimators should verify a uniform COR across the full FOV and collimators with a variability from the mean COR >0.5 pixels should be rejected.

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Optimal performance of a tomographic gamma camera-gantry-computer imaging system requires exact alignment between the electronic, or computer digital image matrix, and the mechanical axes of rotation (AORs) (*1–6*). Misalignment results in artifacts, image degradation, and erroneous values when quantitative analysis is performed. Electronic alignment is dependent upon making the appropriate adjustment of the x and y voltage offsets and gains of the positional amplifier to place the center of the camera crystal at the center of the computer matrix. Mechanical alignment is more complicated and requires a level system base, parallel alignment of the detector head and the axis of rotation, absence of sag or excessive flexibility of the gantry, and perpendicular alignment between the collimator holes and the collimator face for parallel hole collimators (*1,8,9*). If the mechanical and electronic AORs are aligned, then a single center of rotation (COR) measurement is applicable for the entire field of view (FOV). However, verification of alignment requires a more elaborate testing procedure than the routinely performed single point source calculation of the COR. We first defined the maximal acceptable pixel error in the COR that produced artifact-free reconstructed images. We next developed a technique for verifying that potential slices to be reconstructed across the collimator face had COR values within these acceptable limits. This technique was used to test four different collimators.

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METHODS

All studies were performed with a commercially available 400 mm FOV, 61 photomultiplier tube tomographic gamma camera imaging system. Mechanical and electronic components of the system had been appropriately tested according to the manufacturer's specifications. Additional tests were made on the system according to our previously published recommendation (*1*). During the period of collimator testing there were no mechanical or electronic changes made in the system.

COR Error and Generation of Artifacts

A single drop of technetium-99m (^{99m}Tc) was placed into a 2 × 3 mm well in a lucite rod and this point source was placed in the center of the FOV on the AOR, as shown in Figure 1, and an anterior 180° acquisition, consisting of 64 views of 10 sec each and containing 5,000 to 7,500 counts per view, was performed using a 64 × 64 imaging matrix. The 180° acqui-

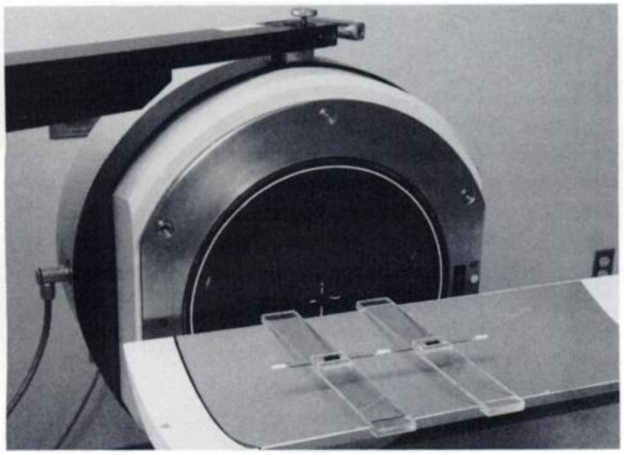

FIGURE 1
The three-point source phantom used in our laboratory for routine COR calculations. The phantom is positioned and held on the axis of rotation by the lucite holders and the table position recorded. This system is easily set up and placement is highly reproducible. For these experiments only a single point contained radioactivity, but normally all three are filled and a single acquisition gives three estimates of the COR; the mean value for the three is used as the COR for reconstructing all slices in the field of view.

sition in a 64 × 64 matrix is routinely used for cardiac acquisition in our laboratory and was selected to approximate clinical patient imaging parameters. The images were reconstructed with a ramp filter. The reconstructed images were scaled such that the most negative value was set to zero. The point source slice location was identified as the slice containing the maximal counts and subsequent reconstructions were performed introducing a COR error, Δ, of 0.5, 1.0 and 2.0 pixels relative to a 128 × 128 matrix. An activity profile through the center of each point was generated for each reconstructed slice containing the point source.

COR Measurements Over Full Field

Since imaging the point source off the AOR during 360° acquisition introduces variation due to spatial gamma camera effects and does not allow the edges of the FOV to be examined, we performed all point source acquisition on the AOR. The horizontal component of the AOR was defined by placing the camera head at 0°, positioning the point source on the center of the imaging table, adjusting the table horizontally until the point was centered in the FOV, and recording the horizontal position. Without moving the point source, the camera head was rotated 90°, the table adjusted vertically to center the point source, and the vertical position recorded. All subsequent experiments were performed using this same horizontal and vertical imaging table position. The point source phantom was placed 5 cm from the edge of the field of view on the AOR. Using a 20-cm radius of rotation, a 360-degree acquisition consisting of 32 views, 10 sec each and containing 5,000 to 7,500 counts per view was performed in a 128 × 128 image matrix. The 360° acquisition was used to allow application of the Fourier analysis method described below and the 128 × 128 imaging matrix increased the resolution limits of the system. Immediately upon completion of acquisition, the point source phantom was moved 5 cm on the AOR, thereby positioning it 10 cm from the edge of the field of view, and the acquisition was repeated. A total of seven distinct point source acquisitions along the AOR and across the entire FOV were performed. This series of acquisitions was performed with four collimators: two high resolution (HR-A and HR-B) and two general all purpose collimators (GAP-A and GAP-B). In addition, 14 separate point source acquisitions, each point 2.5 cm apart, were acquired with the GAP-A collimator. Reproducibility and stability of the system was evaluated by performing seven single point source acquisitions with the HR-B collimator at three different times: two acquisitions on the same day and another acquisition 4 days later.

ANALYSIS

Each of the views of a single point source was background subtracted using a background of five counts per pixel to correct for scatter. The x and y projections of the point images were calculated and filtered in the Fourier domain with a third order Butterworth Filter with a frequency of 0.15 cycles/pixel (*10*). This is a low-pass, or smoothing, filter used to reduce the noise in the projections. Other low-pass filters using a similar pass band should give similar results. The locations of the peak x and y filtered projections were then found by parabolic least squares fitting and extrapolations using five points in the neighborhood of the first approximation to these peak locations. The x and y peak locations versus angle were fitted to the first harmonic, which included the appropriate constant and sin and cos terms, to calculate the average locations and amplitudes. The average x location was defined as the COR and the average y location was the slice number for the point. A regression equation was used to determine the slope and intercept for all points for a given collimator. The slope, which is the tangent of the angle between the electronic and mechanical AOR's, was used to measure the extent of axes misalignment (*1*).

RESULTS

The pixel error in the COR that produces tuning fork artifacts during reconstruction of data acquired over 180° in a 64 × 64 computer matrix was determined by introducing increasing error in the true COR (Fig. 2). The introduced error is expressed in pixels in a 128 × 128 matrix and 1 pixel represents 3.0 mm. An activity profile through the center of the point source was generated for each image. With Δ = 0.5 pixels, there is minimal visual distortion of the point source and the activity profile is narrow and uniform before and after the peak. When the COR error is 1 pixel or greater, progressive widening and dispersion of the point source is present in the images and the activity profile reflects these changes. Data acquired in a 64 × 64 computer matrix will demonstrate tuning fork artifacts with an error of 0.5 pixels or greater as expressed relative to a 128 × 128 matrix. This has also been shown to be true for 128 × 128 acquisition data reconstructed into a 64 × 64 matrix (*1*). Hence, 0.5 pixels is the maximal

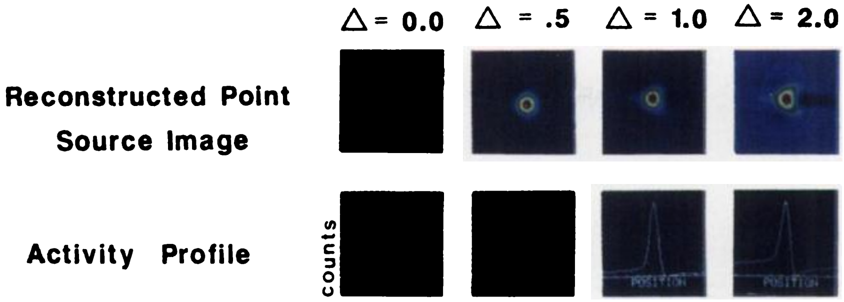

FIGURE 2
The generation of tuning fork artifacts caused by increasing COR pixel error during reconstruction of 180° acquisition in a 128 × 128 computer matrix is evident with errors >0.5 pixels.

allowed COR error. This means that for any collimator the maximal variation from the mean or single measurement COR across the entire collimator face has to be <0.5 pixels in order to have artifact free reconstruction.

Table 1 shows the COR statistics for each of the points measured using four individual collimators. Since the electrical and mechanical components of the imaging system were constant except for the collimator, the observed variability in the mean COR can be attributed to differences due to the collimator itself. There is a maximal COR difference of 1.9 pixels between the mean COR for the GAP-A and HR-A collimators. For individual collimators the COR range for each of the seven points also varies greatly. HR-B had the narrowest variability in the COR for the seven points, 0.47 pixels, and the HR-A had the largest, 1.36 pixels. Thus, the calculated COR for a single point source is dependent upon its location on the collimator face.

The misalignment angle between the electronic and mechanical AOR's ranged from a minimum of 0.3° for HR-B and a maximum of 0.7° for HR-A. Hence, collimators can and do cause apparent mechanical-electrical misalignments.

For each of the four collimators, shown in Figures 3A–D, the individual point source COR is shown as a function of slice on the computer matrix. The mean COR, slope and ±0.5 pixel maximal error limits are also shown. The mean COR varies between collimators and for each collimator there is variability in location related to position on the computer matrix. HR-A and GAP-A have points that are >0.5 pixel from the mean COR and artifacts will be generated in these areas during reconstruction using the mean COR. HR-B and GAP-B have all points falling within the 0.5 pixel error limits and are acceptable for SPECT acquisition (Table 1).

The slice location, calculated COR and the difference from the mean COR for each of 14 points is shown for the GAP-A collimator in Table 2. If the mean COR (64.89) is used for reconstruction, artifacts will be generated in the area of Points 1 and 2. If a single point source COR acquisition were used, for example, points 7 or 8 near the center of the field, artifacts would also be generated in the region of points 3 and 4 since they are 0.5 pixels from the single point COR acquisition (points 7 or 8).

Serial studies were performed with the HR-B collimator at three different time points. The mean COR, slope and intercept were identical for all three acquisitions and indicate the stability of the imaging system and reproducibility of the measurement technique.

TABLE 1
Center of Rotation Statistics Expressed in Pixels Using Four Different Collimators

	GAP-A	GAP-B	HR-A	HR-B
Mean COR	64.89	65.82	66.79	66.24
s.e.m. (±)	0.084	0.013	0.192	0.066
s.d. (±)	0.313	0.273	0.509	0.174
Range	0.99	0.80	1.36	0.47
Maximum difference from mean	0.65	0.48	0.89	0.30

DISCUSSION

Quantitative, artifact free, high resolution SPECT reconstruction of the spatial distribution of radioisotopes over the entire imaging field requires precise alignment between the electronic and the mechanical AORs (*3*). As we have previously described, initial acceptance testing of a SPECT system should verify proper mechanical alignment of the detector gantry, camera head rotation within the gantry, and a level imaging table (*1*). Next, the alignment between the mechanical and electronic AORs should be tested. Tomographic reconstruction in which the COR error for a given slice is >0.5 pixels will lead to image degradation and artifacts.

Our results indicate that the COR is not constant across the FOV during SPECT acquisition and varies from collimator to collimator. Since the electronic and mechanical components of our system were stable, as verified by the reproducibility experiments for a single collimator, the observed variability in COR was due to the collimator. One explanation for this collimator variability is nonperpendicular alignment between the collimator holes and the gamma camera face.

Although several studies have shown the need for perpendicular alignment between the collimator holes

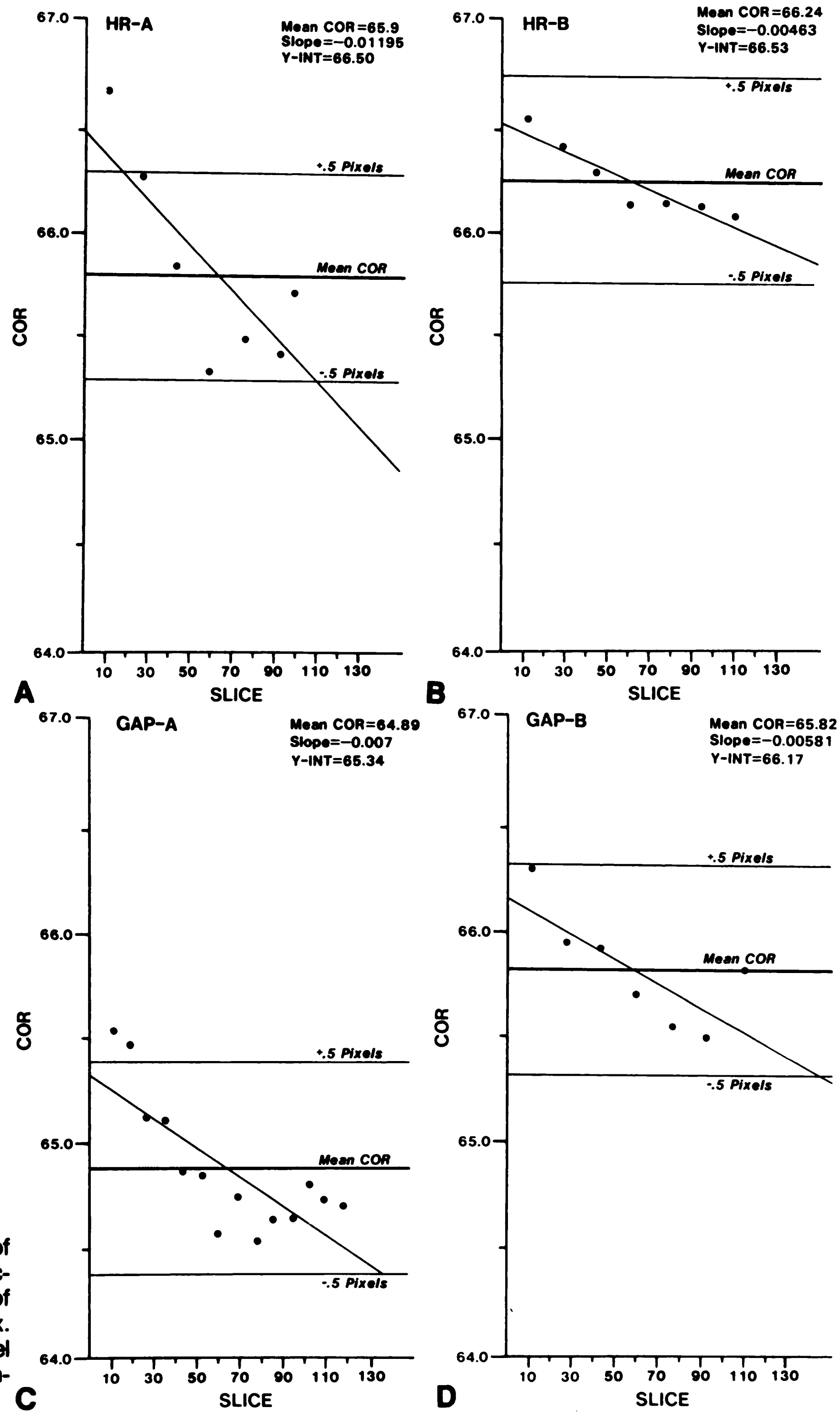

FIGURE 3
A–D: For each collimator the COR of multiple individual point source acquisitions is shown as a function of slice number in the imaging matrix. The mean COR, slope and 0.5 pixel maximal error limits are given for individual collimators.

and face to a tolerance of ±0.50°, clinical users and manufacturers do not consistently perform adequate acceptance testing of collimators (*7–9*). We have shown that image degradation during reconstruction of tomographic data is present when there is a COR error of 0.5 pixels or greater. In the 180° tomographic acquisition, extensively used for cardiovascular imaging, this results in tuning fork artifacts. In 360° acquisition this will result in image degradation and loss of resolution when the COR pixel error is small and generation of ring or annular artifacts when greater error is present.

In the four collimators we tested, two had sufficient variability in the COR over the entire FOV to make them unacceptable for SPECT acquisition. If precautions were taken to avoid locating the organ of interest in those portions of the field having >0.5 pixel error for the mean COR, then these collimators could be used for performing clinical imaging. However, we feel this

TABLE 2
Slice Location, COR, and Difference from the Mean COR for 14 Separate Point Source Acquisitions Using the GAP-A Collimator

Point	Slice location	COR	Difference from mean COR
1	11	65.54	−0.65
2	18	65.47	−0.58
3	26	65.13	−0.24
4	35	65.11	−0.22
5	43	64.87	−0.02
6	52	64.85	−0.04
7	59	64.58	+0.31
8	69	64.75	+0.14
9	77	64.55	+0.34
10	85	64.65	+0.24
11	94	64.67	+0.22
12	102	64.81	+0.08
13	110	64.74	+0.15
14	118	64.71	+0.18

limitation is unacceptable and these two collimators were rejected.

An additional collimator related problem for SPECT imaging may be encountered. Spatial linearity correction, used to correct for the intrinsic spatial distortion of the detector, is performed by some manufacturers using two acquisitions of a precision slit phantom and these corrections became constant for each camera. If this single factory acquisition is performed with a collimator mounted on the detector, the spatial linearity correction may not be accurate when a different collimator is used during clinical imaging.

Thus, after all components of the SPECT camera-gantry-computer system have been tested and have stabilized, multiple point source COR measurements across the entire FOV should be made on each collimator that is to be used for tomography. Once these initial measurements have been made and the collimator found to have <0.5 pixel variation from the mean CDR across the entire FOV, subsequent quality control should ideally include multiple point COR measurements using fewer points. For example, in our laboratory we use three points near the axis of rotation, one point in the center of the field, and the other two ~5 cm from the edges of the field of view. The mean COR for the three points is used during reconstruction of all slices.

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