Dual-Isotope Motion Correction Technique for Gated Exercise Scintigraphy

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In exercise multigated blood-pool imaging, significant degradation of image quality occurs as a result of patient movement under the gamma camera. Motion correction devices using centroid tracking of x-y events emanating from the organ of interest cannot be applied to blood-pool studies, because cardiac contraction and rotation masks the correctable patient motion component. We have developed a dual-isotope motion correction technique (DIMC) which utilizes a second point source of dissimilar energy (^{241}Am) to monitor movement. Positional centroids from events incident in the ^{241}Am window are used to develop correction coordinates which are applied to the ^{99m}Tc blood-pool events. The ability of DIMC to reduce blur due to motion has been evaluated qualitatively with phantoms and quantitatively by using spatial resolution measurements obtained from stationary line sources and from sources moving at varying rates. Based on these criteria, we have found the device to be capable of reducing over 90% of the image blur of objects moving at 5.1 cm per sec. In preliminary gated exercise studies, subjective perception of image quality was shown to be significantly improved in the DIMC corrected image, when compared to images obtained without DIMC. Improvement in image quality for exercise gated studies is of particular importance because of the low count density obtained during these procedures.

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Exercise gated blood-pool studies are useful to assess coronary artery disease when changes in ejection fraction and in regional wall motion at peak exercise are evaluated (*1,2*). Further, we among others have shown that the skewness of phase histograms generated during exercise can be a useful indicator of coronary artery disease when peak exercise studies are analyzed by phase and amplitude analysis (*3,4*). Exercise gated studies are hampered by low statistics due to the relatively short acquisition periods, and are degraded significantly by patient movement under the gamma camera during vigorous exercise. To accurately assess indicies of left ventricular function from these studies, precise definition of the left ventricular chamber is essential (*5*).

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As with any imaging method, movement of the organ of interest during acquisition reduces image quality by introducing a blur function. Blur may be due to various types of patient or organ motion, including patient respiration. Oppenheim and Hoffer et al. (*6-8*) have shown that motion blur due to respiration or patient movement is correctable by activity centroid tracking, as changes in the centroids of activity in the x-y axes are due to positional shifts of the organ. Thus, planar motion of an organ assumed to be a rigid body, such as the liver, is correctable (*9*). In multigated blood-pool imaging, blur occurs both as a result of patient motion and intrinsic contraction of the heart; the latter results in organ deformation, rotation, and motion perpendicular to the image plane, which are not correctable by centroid tracking. During exercise studies, patient motion blur is severe, yet the ability to compensate for this correctable motion component would be impaired by cardiac contraction, if technetium-99m (^{99m}Tc) blood-pool activity centroids were utilized.

To overcome the inability of centroid tracking methods

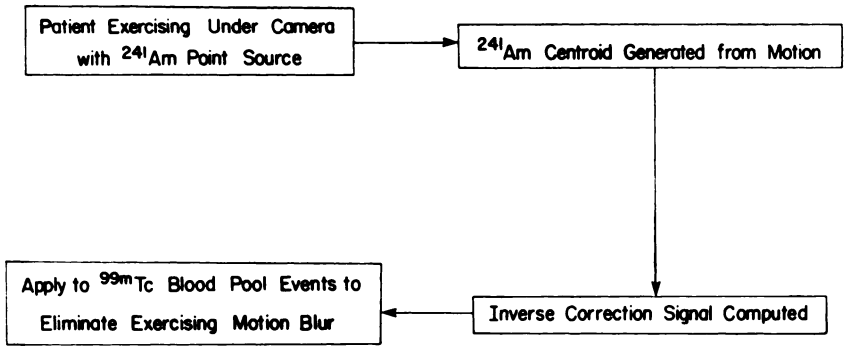

FIGURE 1
Principle of dual-isotope motion correction. ^{241}Am is taped to patient's chest and provides reference source to monitor motion. Correction signals are applied to ^{99m}Tc blood-pool events

to correct for patient movement utilizing ^{99m}Tc events in gated blood-pool studies, a dual-isotope motion correction device (DIMC) was developed. The device uses a second point source of dissimilar energy, the 60 keV x-ray of americium-241 (^{241}Am), to monitor patient movement under the gamma camera. This report describes the DIMC principle, evaluation, and initial clinical trial.

MATERIALS AND METHODS

The dual-isotope motion correction device consists of an analog circuit which continuously computes the x-y position centroids from reference source events which fall in the first of two gamma camera energy windows. For this study, a 17.1 mCi ^{241}Am point source was utilized for reference, with a 20% energy window centered at 60 keV. The DIMC, integrated within the gamma camera, develops x-y activity centroids by continually summing the $\pm x$ and $\pm y$ position signals from valid photopeak events from the reference source window. The net x, y coordinate represents the centroid of the incident activity distribution. After "locking" on the initial activity centroid location (e.g., $x = -1.0$ V, $y = +0.6$ V), subsequent shifts in the x, y centroids are continuously determined by comparing the current centroid location to the initial centroid location. Corrected x-y signals are generated simply as x-y offset voltages representing the magnitude of x-y centroid excursion from the initial location, but having opposite sign. In normal operation of a "Hoffer type" motion correction circuit, these correction signals would be applied to the events incident in the same window. However, the DIMC is designed to apply these correction (voltage) signals to photopeak events falling in a second analyzer window, that used for ^{99m}Tc events. A real time correction is applied to each ^{99m}Tc photopeak event based on the current centroid location. Thus, the x-y outputs from the gamma camera to the computer represent valid photopeak, motion corrected ^{99m}Tc blood-pool events. A schematic representation of the DIMC concept is given in Fig. 1.

Movement of the point source centroids spatially follows patient movement under the gamma camera; this component is also present in the ^{99m}Tc blood-pool events. Correction signals applied to the ^{99m}Tc blood-pool events generated by the motion tracking point source centroids effectively remove the motion blur from the blood-pool image as long as both the point source and cardiac blood pool both remain within the gamma camera field of view. The 60 keV analyzer window is sufficiently removed from the ^{99m}Tc photopeak so that the patient scatter contribution in the 60 keV window is relatively small in relation to the number of photopeak events from the Am-241 source.

The characteristics of the DIMC have been studied through phantom experiments and quantitative assessment of spatial resolution. Further, the device has initially been tried on patients undergoing the typical stress gated cardiac imaging protocol at our institution.

Typically, error in the determination of centroid location produces a loss of spatial resolution, when motion correction devices are utilized to image stationary objects. Errors in centroid location result due to the finite number of events sampled per unit time and the geometry of the ^{241}Am centroid source. For the DIMC this effect is compounded by the fact that technetium scattered events incident in the 60 keV window will contribute to an error in the centroid location. To study the loss of resolution using DIMC, system resolution was obtained using a distributed and a single point ^{241}Am reference source, at varying ratios of distributed ^{99m}Tc scatter to ^{241}Am photopeak events. To provide a distributed americium source, multiple point ^{241}Am sources were used across the camera field of view. Count rates from the ^{241}Am source were attenuated with a graded set of copper absorbers and activity varied to produce different ratios of ^{241}Am events to ^{99m}Tc scattered events in the 60 keV window. Technetium scattered events were obtained from a ^{99m}Tc-filled phantom.

To qualitatively study the ability of DIMC to improve spatial resolution, images were obtained from a Rollo phantom moved at an average rate of 4 cm per sec. Rollo phantom images were obtained with no motion; with motion at 4 cm per sec, uncorrected; and with motion correction using DIMC. For quantitative evaluation of DIMC performance, full width at half maximum (FWHM) gamma camera system resolution, using a previously described method (*10*), was determined from stationary and moving line sources at a distance of 2 cm from a low-energy, all-purpose (LEAP) collimator.

To test motion correction, an apparatus was built to move ^{99m}Tc line sources in front of a portable gamma camera at varying speeds up to 5.1 cm per sec. This apparatus consisted of a digital stepping motor capable of driving a platform containing three ^{99m}Tc line sources (1mm in diameter) placed 30 cm apart, and collimated ^{241}Am reference sources. A digital counter allowed precise adjustment of the speed of the motion. Spatial resolution was obtained from an average of the FWHM of the three line spread functions (LSF). Performance of the device was determined at various motion rates and reference source (^{241}Am) strengths. Physical linear excursion of the line sources was fixed at 40 mm for all motion rates. Precision of

TABLE 1
Noise Equivalent Motion Blur Due to ^{99m}Tc Scatter

Ratio of ^{241}Am to ^{99m}Tc scatter	DIMC off (mm)	DIMC on (mm)	Δ (mm)
24:1	5.1	5.3	0.2 ± 0.3
12:1	5.1	5.3	0.2
8:1	5.1	5.7	0.6
4:1	5.1	5.8	0.7

line and point source motion "tracking" and accuracy of motion correction were evaluated.

Patient studies

During supine exercise, a reference point source consisting of 17.1 mCi of ^{241}Am was taped to the patient's chest in a location which would not interfere with events emanating from the cardiac blood pool. For the americium point source, a special tantalum housing was developed to (a) shield, thus reducing the radiation burden to the patient, and (b) to provide collimation, restricting the americium events onto a relatively small area on the gamma camera. The tantalum housing contained a 5-mm-deep conical aperture, tapered at 60° from the central axis, with an exit aperture for the ^{241}Am events 2 mm in diameter at its base. The 60° taper allows an adequate flux of ^{241}Am events onto the collimated camera, with the sources placed on the patient with up to a 45° tilt to the camera plane. The 45° tolerance provides adequate leeway to position the reference source on the patient in appropriate locations, with acceptable ^{241}Am flux, while maintaining reasonable collimation.

Preliminary clinical trials were performed on two patients and one normal volunteer who signed informed consent under the guidelines of our institution's Human Investigation Committee and underwent exercise gated scintigraphy with and without the dual-isotope motion correction technique. The DIMC circuit, installed on a conventional mobile gamma camera*, provided simultaneous corrected and uncorrected outputs so patients did not need to undergo the exercise protocol twice. Corrected and uncorrected data were acquired by two identical microcomputer systems† and improvement in image quality was evaluated subjectively by presentation to trained observers the corrected and uncorrected cine mode display of blood-pool images at peak exercise, for this initial evaluation.

RESULTS

The loss of resolution or noise equivalent blur due to the source distribution was evaluated using the FWHM of LSFs 2 cm from the LEAP collimator. Without DIMC, the system resolution was measured to be 5.1 ± 0.3 mm from stationary ^{99m}Tc line sources. When the DIMC was activated, the system resolution with a point source of ^{241}Am in place was 5.3 ± 0.3 mm (virtually identical to the system resolution obtained with the DIMC off). However, a system resolution of 5.9 mm ± 0.3 mm was obtained (a loss of 0.8 mm ± 0.3 mm) with the DIMC activated using spatially distributed americium sources. Thus, essentially no loss in spatial resolution was seen when using the DIMC for a *point source* of activity since, for a very tightly grouped set of positional coordinates, very little error is introduced in computing the centroids. If no ^{99m}Tc scatter was present, virtually no loss in system spatial resolution was introduced by the DIMC for reference point source strengths from 4,000 to 12,000 counts per sec. However, when technetium scatter was present, then noise blur due to a distributed *technetium* scatter source was seen to vary with the ratio of ^{241}Am events to ^{99m}Tc scatter. The noise equivalent blur due to technetium scatter in the americium window is shown in Table 1, for ratios of americium activity to technetium scatter of 4:1 to 24:1. As can be seen, for ratios of americium activity to technetium scatter of 24:1 virtually no resolution is lost through the DIMC. However, at ratios of 8:1 or less, the loss of resolution in motion studies employing the DIMC equates to ~ 12%

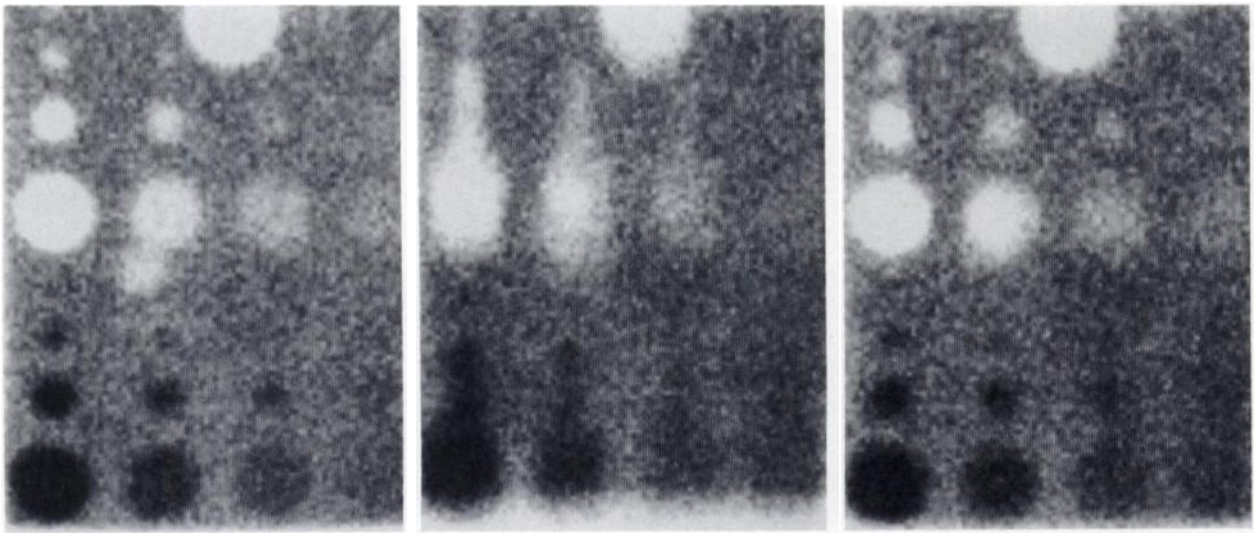

FIGURE 2
Images of Rollo phantom obtained without motion (left), uncorrected motion at 4 cm/sec. (center) and DIMC corrected (right). Cold (white) circular area at top is ^{241}Am holder. Extra cold area seen in "no motion" image is air bubble left in phantom. Air bubble is not seen in uncorrected/corrected images as its effect is averaged out by motion

TABLE 2
Spatial Resolution* Compared with Rate of Motion (Speed)†

Speed (cm/sec)	FWHM (mm)	Δ (mm)
0	5.1 ± 0.3	– ± 0.3
2.0	5.7	0.6
2.7	6.0	0.9
3.3	6.5	1.4
4.0	7.1	2.0
5.1	7.8	2.7

*FWHM ^{99m}Tc LSF 2cm from face.
†Using 12 K/sec ^{241}Am with LEAP collimator for 40.0mm excursion.

(purely due to distributed ^{99m}Tc scatter in the reference window).

The results obtained from Rollo phantom images moved at 4cm per sec are shown in Fig. 2. The additional cold spot present at the top of the Rollo phantom is due to the shielded americium reference point source placed on the Rollo phantom. Note the improvement in image quality when the DIMC was employed. The image of the DIMC corrected moving Rollo phantom (far right) appears virtually identical to phantom images obtained with no motion (far left). The additional small cold spot in the Rollo phantom, visible in the study obtained with no motion, was due to a bubble left in the phantom. Although the DIMC corrected image appears similar to the "no motion" image, the bubble cold spot is no longer visible as its effects were averaged out by the motion of the phantom.

The FWHM system resolution using a 12,000 count per second (cps) ^{241}Am reference source and ^{99m}Tc scatter at speeds from 0 to 5.1 cm per sec are shown in Table 2. For this experiment, each line source experienced a 40 mm physical displacement at various velocities. Images of the uncorrected and corrected line sources are provided in Fig. 3. At speeds up to 2.0 cm per sec the FWHM of the line spread function was approximately the same as would be expected from the noise equivalent blur due to technetium scatter within the americium window. At 5.1 cm per sec the FWHM was 7.8 mm (an increase of 2.7mm over the stationary spatial resolution). The total excursion, however, was 40 mm thus, the DIMC was able to correct for 93% (37.3 mm/40 mm) of the physical excursion of the line sources.

Since the DIMC updates the motion centroid after ~ 1,000 events have been accumulated, higher americium count rates will produce an improved response of the system at higher motion rates. Since larger count rates from the americium source will increase gamma camera deadtime it becomes a practical problem to determine the optimum Americium activity. When the velocities of correctable motion were evaluated as a function of count rate of the reference source, the FWHM was not significantly increased when DIMC was activated with no ^{99m}Tc scatter in the absence of motion (velocity = 0) for count rates from 4,000 to 12,000 counts for a point source. As the line sources were moved at velocities of 2 cm per sec to 5.1 cm per sec, it was readily apparent that the FWHM increased significantly at motion rates of 5.1 cm per sec as the reference source strength decreased from 12,000 CPS to 4,000 CPS. System resolution compared with source strength for motion velocities of 2 cm per sec and 5.1 cm per sec are shown in Table 3.

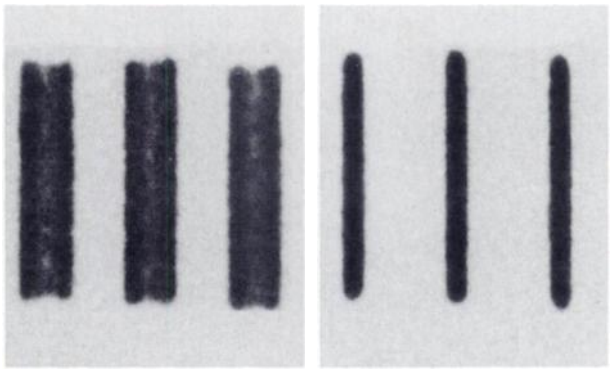

FIGURE 3
Images of line sources moved at 5.1 cm/sec. (40 mm total excursion), uncorrected (left), corrected (right). Uncorrected line sources show gross motion blur with pattern reflecting back and forth (simple harmonic) motion

End diastolic and end systolic images obtained from a normal patient at maximal exercise with and without DIMC are shown in Fig. 4. Note the improvement in image quality for the motion corrected images, particularly in the end systolic frame. In the uncorrected end systolic image the margins of the left ventricle are extremely ill defined, particularly in the area of the intraventricular septum.

DISCUSSION

Noise blur is introduced into an image when any motion correcting circuit is activated due to statistical uncertainty in locating the events in the finitely sampled distribution used for centroid computation. Since a finite number of events are sampled when the centroid is computed, there will be a statistical "noise" component introduced by the circuit. In the case of the dual isotope motion correction device, a relatively tight distribution of activity is produced by the (moving) point source, with a relatively small "circle of uncertainty". Less noise is introduced into the circuit using a point reference source than from centroids computed from a distributed activity source,

TABLE 3
Spatial Resolution* Compared with Reference Source Strength

	^{241}Am source strength			
SPEED (cm/sec)	12K	8K	6K	4K (/sec)
0 DIMC off	5.1	5.1	5.1	5.1 ± 0.3mm
0 DIMC on	5.3	5.4	5.4	5.4
2.0	5.7	6.2	7.0	9.0
5.1	7.8	9.9	13.7	16.7

*FWHM ^{99m}Tc LSF 2cm from LEAP collimator.

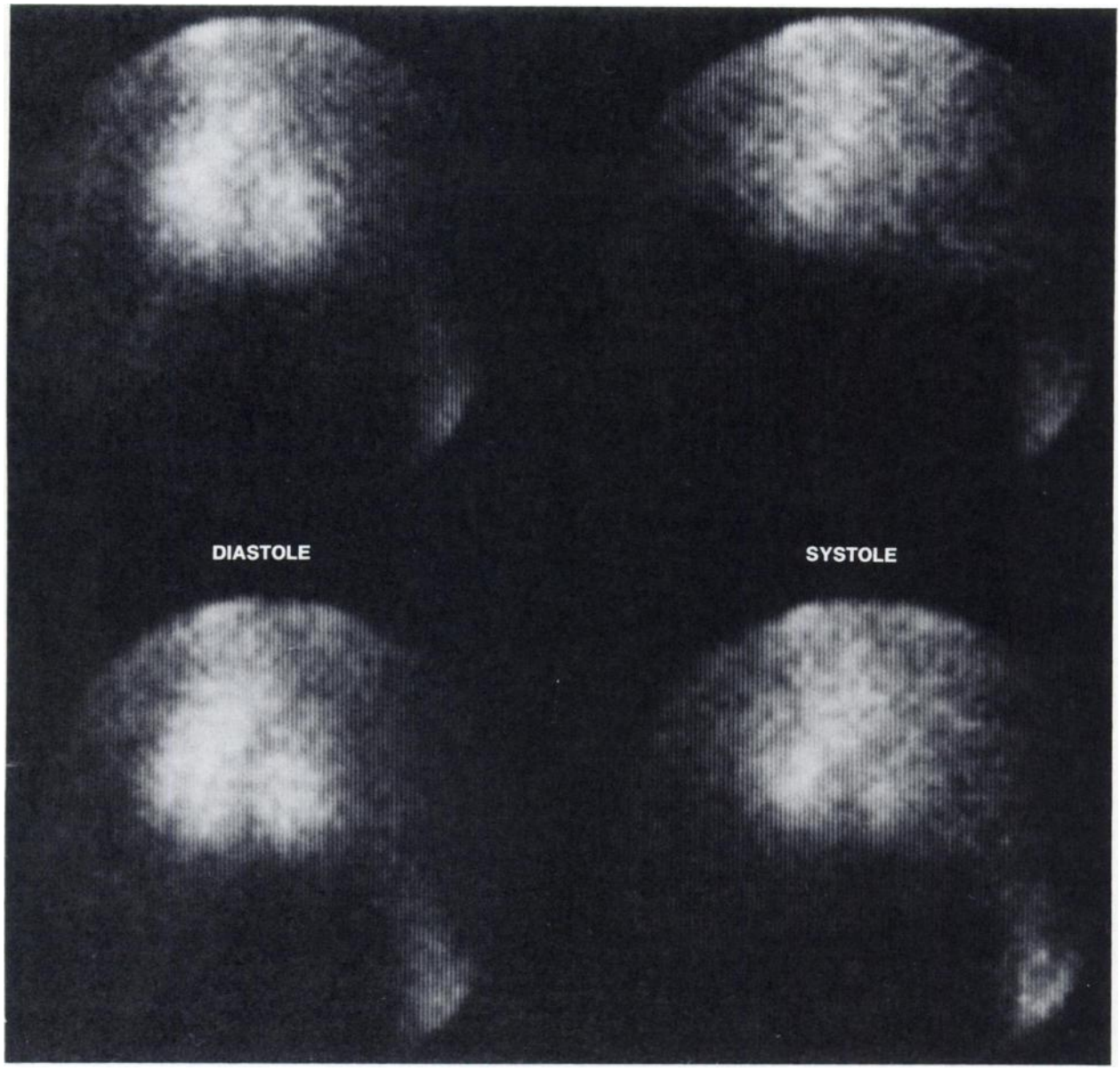

FIGURE 4
End-diastolic/end-systolic LAO views from stress gated study performed on normal patient at peak exercise (100W), uncorrected (top), and DIMC corrected (bottom). Note improved image quality, particularly in end-systolic image

such as those used in conventional motion corrected liver studies.

With DIMC, technetium scattered events within the 60 keV window, will not only produce noise blur due to the spatially distributed "second window" activity, but will also introduce error in the centroid calculation if these scattered events emanate from moving parts of the patient, as they would represent non-correctable components of cardiac motion, e.g., cardiac contraction. Using a reference source with high (over 140 keV) energy emissions would eliminate resolution loss due to ^{99m}Tc scatter, but would introduce problems of collimation and degrade image quality by scattering events into the ^{99m}Tc window. It is imperative that an ^{241}Am source strength be great enough to provide at least a 4:1 ratio of americium events to technetium scattered events. It is logical to assume that the tight spatial distribution of ^{241}Am 60 keV events will significantly bias the centroid, more so than the distributed ^{99m}Tc scatter which will tend, by spatial averaging, to cancel contributions to the x and y centroids. The net result should be an improved accuracy in centroid location, for studies with motion.

Obviously, increasing the Am-Tc ratio above 4 or 5:1 would have a further beneficial effect on reducing noise blur from the device, more importantly, since the device samples ~ 1,000 events before it updates centroid location, larger motion rates could be accommodated. To increase ^{241}Am source strength, however would be prohibi-

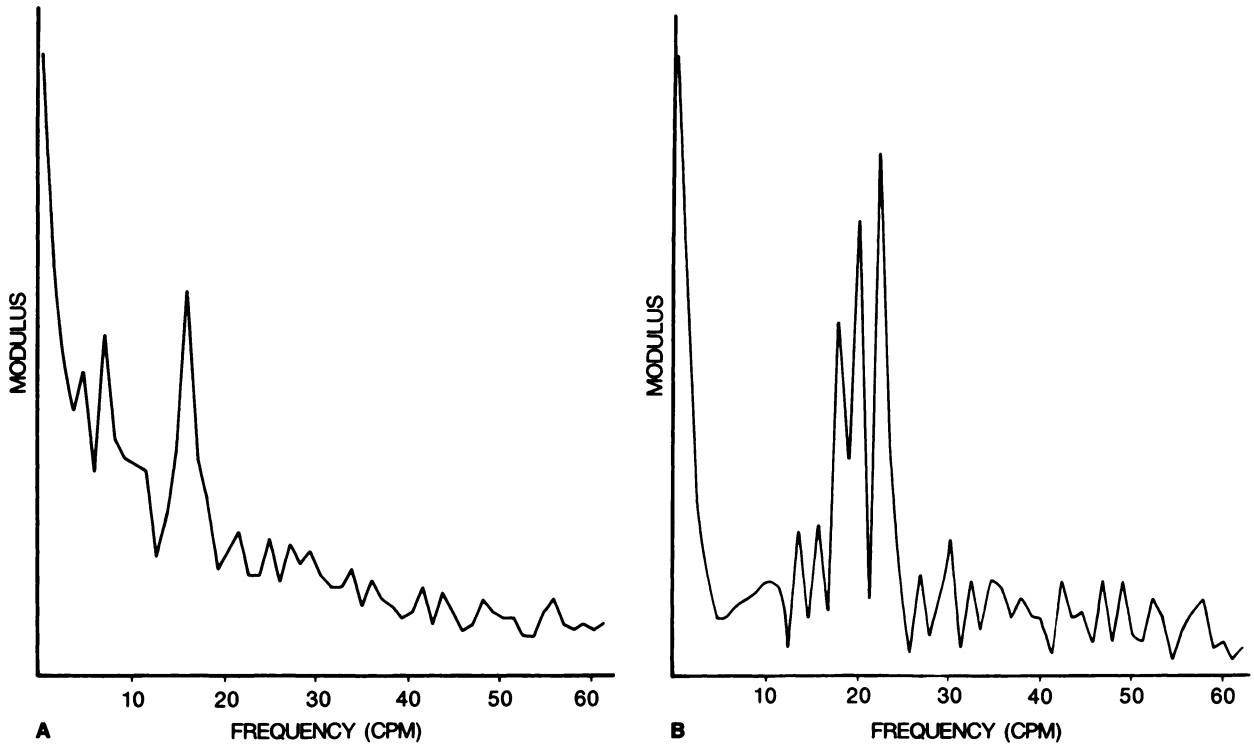

FIGURE 5
A: Modulus of Fourier transform (ordinate) of "y" motion centroid ("y" = patient long axis) vs. frequency (abcissa) in cycles per min (cpm) for resting gated blood-pool study. Note peak in Fourier spectrum at 17 cpm from respiration (RESP). B: Modulus of Fourier translation of "y" motion centroid vs. frequency from a [^{99m}Tc]SC liver study. Pronounced peak at 20 cpm from RESP indicating much larger centroid excursion with RESP in liver studies. Gated blood-pool RESP centroid excursion (A) masked due to complex cardiac motion, yet discernable component present. B: Multiple peaks observed near 20 cpm are artifactual due to Fourier transform routine (due to finite temporal sampling of centroid and Poisson noise)

tive due to increased gamma camera deadtime for most gamma cameras without high count rate capability. We have found that reference source strengths of ~ 12,000 cps provide an optimum activity for motion correction, without significantly prolonging gamma camera deadtime in typical gated blood-pool studies. Twelve thousand CPS reference source strength provides approximately a 4:1 ratio of ^{241}Am events to patient ^{99m}Tc scattered events in a 20% 60 keV window for a typical patient.

In devices of this type, it is not appropriate to specify the magnitude of motion which can be compensated for, since theoretically, if the motion rate is slow enough, an infinite displacement of an object could be corrected. It is, therfore, only appropriate to define the characteristics of the device in terms of the *rate* of correctable motion, i.e., how well is organ motion effectively removed as the velocity of motion increases. The correctable velocity is dependent primarily on the count rate from the reference source and to a lesser degree on the collimator being used for the study (as the latter affects camera sensitivity).

Our initial study of motion degradation in gated cardiac imaging has been previously reported (*11*). Although respiratory motion in gated cardiac imaging is estimated to be of the order of 1 to 2 cm, use of a conventional motion correction device for blood-pool studies results in a distorted cardiac silhouette. Fourier analysis of motion centroids from the cardiac blood pool revealed a peak at the respiratory rate in the y direction, defined as the long axis of the patient. As shown in Fig. 5A, this peak is significant, however, not as significant as the respiratory peak obtained from a liver study shown in Fig. 5B. One would suspect that regular motion during exercise would result in a similar peak in the Fourier spectrum of the activity centroids. Perhaps a filter could be applied retrospectively to the data set at the appropriate frequency to remove the effects of respiration and/or periodic patient motion exhibited during exercise. We are currently investigating this approach.

Wedel et al. (*12*) has attempted to reduce motion blur in exercise studies using a complicated restraining harness. Use of DIMC would eliminate the need for extraordinary patient immobilization, allowing faster patient set up and improved comfort. Utilization of DIMC may also be appropriate in non-exercise studies where the patient is uncooperative or moving excessively during gated cardiac acquisition. Potentially, DIMC can be used for treadmill gated exercise studies with the patient standing.

In practical clinical nuclear medicine, our initial use of DIMC indicates that it provides a convenient real-time method to correct for motion in exercise gated cardiac

studies. Currently, an extensive clinical trial is underway to quantitatively evaluate improvements in image quality in the use of DIMC when evaluating quantitative parametric information such as ejection fraction and phase histograms.

In conclusion, initial studies indicate a significant improvement in exercise gated blood-pool images is obtained using DIMC. Possibly the most important application of the device is in elderly or out of condition patients who move more vigorously under the gamma camera in an effort to keep up with the increasing workload of an exercise protocol (these patients are often most in need of a "motion free" accurate study). Any translation of the patient's heart (and body) under the gamma camera can be corrected provided that the heart as well as the ^{241}Am point source remain within the gamma camera field of view. Thus the use of large field of view cameras may allow more latitude in patient movement under the gamma camera, as the heart and reference source are imaged using a wider area of view. The DIMC provides a commercially available‡, real-time method to correct for motion in exercise gated studies.

FOOTNOTES

*Siemens Medical Systems, Iselin, NJ—LEM Mobile Camera.

†MicroDELTA computer.,

‡Siemens Centroid Source Motion Corrector.

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