cost and public concern. The relation of reduced coronary flow reserve and hypercholesterolemia is a current topic in the field of clinical cardiology. The ability to measure MBF in the rabbit model, in particular the hyperlipidemic rabbit, could be important in this field.

CONCLUSION
We have demonstrated the capabilities of our cardiac PET system for in vivo imaging and quantitation of MBF in small animals. This system has sufficiently high temporal and spatial resolutions to be used on small animals and may have the potential to play an important role in the study of animal models in cardiovascular diseases.

ACKNOWLEDGMENTS
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REFERENCES

Comparison of Four Motion Correction Techniques in SPECT Imaging of the Heart: A Cardiac Phantom Study

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The aim of this study was to evaluate the accuracy of four different motion correction techniques in SPECT imaging of the heart.

Methods: We evaluated three automated techniques: the cross-correlation (CC) method, diverging squares (DS) method and two-dimensional fit method and one manual shift technique (MS) using a cardiac phantom. The phantom was filled with organ concentrations of $^{99m}$Tc closely matching those seen in patient studies. The phantom was placed on a small sliding platform connected to a computer-controlled stepping motor. Linear, random, sinusoidal and bounce motions of magnitude up to 2 cm in the axial direction were simulated. Both single- and dual-detector 90° acquisitions were acquired using a dual 90° detector system. Data were acquired over 180° with 30 or 15 frames/detector (single-/dual-head) at 30 sec/frame in a 64 × 64 matrix. Results: The simulated single-detector system, CC method, failed to accurately correct for any of the simulated motions. The DS technique overestimated the magnitude of phantom motion, particularly for images acquired between 45° left anterior oblique and 45° left posterior oblique. The two-dimensional and MS techniques accurately corrected for motion. The simulated dual 90° detector system, CC method, only partially tracked random or bounce cardiac motion and failed to detect sinusoidal motion. The DS technique overestimated motion in the latter half of the study. Both the two-dimensional and MS techniques provided superior tracking, although no technique was able to accurately track the rapid changes in cardiac location simulated in the random motion study. Average absolute differences between true and calculated position of the heart on single- and dual 90°-detectors were 1.7 mm and 1.5 mm for the two-dimensional and MS techniques, respectively. The corresponding values for the DS and CC techniques were 5.7 and 8.9 mm, respectively. Conclusion: Of the four techniques evaluated, manual correction by an experienced technologist proved to be the most accurate, although results were not significantly different from those observed with the two-dimensional method. Both techniques accurately determined cardiac location and permitted artifact-free reconstruction of the simulated cardiac studies.

Key Words: motion correction; SPECT; myocardium


MOTION CORRECTION IN CARDIAC SPECT • O'Connor et al. 2027
Tomographic imaging of the heart with either $^{201}$Tl or $^{99m}$Tc-based radiopharmaceuticals is one of the most frequently performed procedures in nuclear medicine. All tomographic studies require that the object being imaged remain constant, both in terms of its spatial location and distribution of activity, for the duration of the tomographic acquisition. Movement of the object may result in artifactual image reconstruction, depending on the magnitude and direction of motion. While motion artifacts are not considered to be a significant problem in general tomographic imaging, they may affect a significant percentage of studies in tomographic imaging of the heart. This is because the aftereffects of stress testing and the requirement of the patient to keep their arms above their head increase the likelihood of motion artifact. The advent of dual-head and triple-head SPECT systems has considerably shortened the time required to acquire a myocardial SPECT study and should, in theory, also reduce the incidence and severity of motion artifacts. However, our experience of dual-head SPECT systems at the Mayo Clinic has shown that these systems have increased the perception of patient motion in clinical studies. This is largely due to the fact that a review of the raw planar data in cine format will often show a shift in cardiac position between the last frame acquired on Head 1 and the first frame acquired on Head 2. While these two frames are contiguous in terms of their spatial representation of the heart, they are acquired at either ends of the acquisition. Any minor patient motion over the duration of the study will appear as a sudden shift in cardiac position between these two frames. Dual- and triple-head gamma camera systems have enhanced the ability of the user to detect patient motion and increased the need for accurate motion correction techniques. At the Mayo Clinic, we estimate that approximately 10%–20% of studies contain sufficient motion to require application of motion correction software or repeat acquisition of the patient study. Motion correction is preferable to repeating the acquisition, particularly for $^{201}$Tl where redistribution can occur between acquisitions.

Several different acquisition and analysis techniques have been proposed to reduce the severity of motion artifacts. Some of these techniques are applicable only to certain types of detector systems, such as the "temporal image fractionation" technique described by Germano et al. (1), which can be used on a triple-head or dual-head 90° detector system. Most techniques attempt to correct for patient motion by suitable shifting of the image data in the axial direction. The main obstacle facing these techniques is accurate identification of the myocardium in the planar data. Some techniques overcome this problem through the use of an external marker (2,3) but are then limited in their ability to correct for heart movement within the chest cavity, such as occurs with myocardial creep (4). The most widely used correction techniques use various algorithms to detect the position of the heart in successive frames and compute axial shifts of these images to eliminate apparent cardiac motion (5–7). These are the cross-correlation (CC) technique (5), diverging squares (DS) method (6) and two-dimensional fit technique (7). All three techniques are available on many commercial nuclear medicine computer systems.

The CC technique is based on a discrete cross-correlation function between two one-dimensional sequences of data (5). To create the one-dimensional sequence, the summed profile along the y-axis is created for each image. Profiles from successive images are compared using a cross-correlation function, which can determine the magnitude of the shift in the distribution of profile counts from one image to the next. Since this technique uses the entire y-profile, it is sensitive to the presence of noncardiac activity.

The DS technique requires the user to first detect the myocardium by placing a small box in the middle of the heart (6). The algorithm then increases the size of the square box by including those adjacent rows and columns that maximize the total counts in the new box. This process is repeated until a final box size of $\sim 6 \times 6$ cm is created over the heart. The center of this box is considered the center of the heart. For successive projections, the algorithm starts with the center of the heart from the preceding projection and follows the above algorithm to create a new box. Unlike the CC technique, this method will not be influenced by activity outside the box but will be sensitive to overlapping hepatic or bowel activity.

The two-dimensional fit method requires the user to define a circle that encompasses the heart on the 45° left anterior oblique projection (7). In a manner somewhat analogous to the CC technique, the two-dimensional method attempts to minimize the differences in the spatial distribution of activity inside the circle between adjacent projections. Like the DS technique, this method will be unaffected by activity outside the circle.

Previous studies examining the effectiveness of automated motion correction algorithms have relied on the artificial generation of motion in the tomographic data either by computer manipulation of the image data (5,7) or by an abrupt manual shift of the imaging table during the acquisition (3,6). While these can permit an evaluation of the effectiveness of a correction algorithm, they do not fully model all the effects of patient motion (e.g., smearing of the image data that occurs with myocardial creep) and can give a misleading impression of the ability of the algorithm to correct for patient motion. Furthermore, studies of the CC, DS and two-dimensional techniques have evaluated their effectiveness in $^{201}$Tl studies (3–7). The different biodistributions of $^{201}$Tl- and $^{99m}$Tc-based radiopharmaceuticals may alter the ability of these algorithms to accurately detect myocardial motion.

The aim of this study was to compare the effectiveness of three automated methods of motion correction in $^{99m}$Tc-sestamibi cardiac SPECT studies using a cardiac phantom attached to a special, motion-controlled platform. In addition to the three automated techniques, we also evaluated a simple correction technique based on the manual shift (MS) of the images in the axial direction.

**MATERIALS AND METHODS**

**Myocardial Phantom**

Phantom studies of the heart were obtained using an anthropomorphic phantom (Data Spectrum, Chapel Hill, NC) containing a myocardial phantom (Model RH-2; Capintec, Inc., Ramsey, NJ). The myocardium compartment was filled with $^{99m}$Tc at a concentration of 278 kBq/ml (7.5 µCi/ml). The background and liver/spleen compartments were filled with $^{99m}$Tc at concentrations of 7.4 kBq/ml (0.2 µCi/ml) and 48 kBq/ml (1.3 µCi/ml), respectively. The lung compartment was filled with polystyrene beads and $^{99m}$Tc at the concentration of 11 kBq/ml (0.3 µCi/ml). These concentrations of $^{99m}$Tc were designed to simulate those seen in clinical studies (8). In addition, to ascertain patient motion effect on the visualization of a small infarct, a small defect (20% of myocardial volume) was placed in the anterolateral wall of the myocardium.

**Motion Simulation**

To simulate patient motion, the cardiac anthropometric was placed on a small sliding platform as in Figure 1. The platform was connected by a shaft to a stepper motor with a resolution of 0.002 mm/step. The speed and direction of the stepper motor was computer controlled and was programmed to perform a variety of linear, sinusoidal and random motions as shown in Table 1 and
single- and dual-head 90° acquisitions with myocardial creep (linear motion), respiratory motion (sinusoidal motion), isolated patient motion (single sinusoidal bounce) and random patient motion (random linear motion), based on the reported frequency and type of motion in clinical studies (9).

**Motion Correction**

Four different motion correction techniques were evaluated. These were the CC technique, the DS technique, the two-dimensional fit technique and MS of the image data.

For evaluation of the CC and DS techniques, data were transferred to a Pinnacle system (Medasys, Ann Arbor, MI) where commercially available implementations of both algorithms were used. Implementation of the CC technique was based on the original description of the algorithm by Eisner et al. (5) without modification. The implementation contained a threshold of 1/4 pixel (1.6 mm) below which motion was not corrected. The implementation of the DS technique was based on the original description of the algorithm by Geckle et al. (6) without modification. No threshold for patient motion was set.

A commercially available motion correction software package was used on an SP computer system (Elscint) to evaluate the MS and two-dimensional techniques. The implementation of the two-dimensional technique was based on the original description of the algorithm by Cooper et al. (7) without modification. The user could define a lower threshold for motion detection. This was set to zero for all studies. In addition, for dual-head 90° acquisitions, the two-dimensional technique was optimized to take advantage of the fact that identical patient motion was present in frames 1–15 and 16–30. Hence, frames 16–18 were used to determine motion in frames 1–3 and frames 13–15 used to determine motion in frames 28–30. For the MS technique, an experienced nuclear medicine technologist corrected all six motion studies with no a priori knowledge of the type or magnitude of motion artifact in the studies. The MS technique displayed a horizontal line across a cine display of the raw data and blanked out those parts of the displayed images above and below the myocardium as shown in Figure 2. The operator could perform axial shift of each image to within 1/5 pixel, coupled with a continuously updated cine display of the corrected planar images.

With all four motion correction techniques, the magnitude of the correction applied to each image was recorded in pixels, converted to centimeters and expressed in terms of the absolute position of the myocardium relative to the first image in each study. Since the phantom was in motion during each acquisition, the true position of the myocardium was obtained by integrating its motion over the duration of each frame and calculating its average position. This calculation took into account movement of the phantom during gantry motion. The absolute differences between the true position

**TABLE 1**

Description of Different Types of Cardiac Motion Simulated in Study

<table>
<thead>
<tr>
<th>Acquisition no.</th>
<th>Acquisition detector heads</th>
<th>Type of axial motion</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>2</td>
<td>None</td>
</tr>
<tr>
<td>2</td>
<td>1</td>
<td>2-cm linear motion over duration of acquisition (1016 sec)</td>
</tr>
<tr>
<td>3</td>
<td>2</td>
<td>2-cm linear motion over duration of acquisition (506 sec)</td>
</tr>
<tr>
<td>4</td>
<td>1</td>
<td>Sinusoidal “bounce,” magnitude 2 cm, duration 240 sec commencing after 450 sec of the 1016-sec acquisition</td>
</tr>
<tr>
<td>5</td>
<td>2</td>
<td>Sinusoidal “bounce,” magnitude 2 cm, duration 240 sec commencing after 200 sec of the 506-sec acquisition</td>
</tr>
<tr>
<td>6</td>
<td>2</td>
<td>Continuous sinusoidal motion, magnitude 2 cm and duration of sinusoidal cycle = 266 sec of the 506-sec acquisition</td>
</tr>
<tr>
<td>7</td>
<td>2</td>
<td>Random linear motion in the range ± 1 cm with change in direction and speed every 50 sec during the 506-sec acquisition</td>
</tr>
</tbody>
</table>
of the myocardium and that estimated by each of the four motion correction techniques were determined. The average values of these differences were obtained for all 30 images in each study and expressed as an average absolute error (in centimeters) per image for each technique. The effect of motion on tomographic image quality was visually assessed from reconstructed long- and short-axis slices of the heart, both pre- and post-motion correction.

RESULTS

Figure 3 illustrates the effects of the different types of cardiac motion on the quality of the short-axis slices of the heart. While the magnitude of cardiac motion was similar in all studies, those simulating linear motion of the heart during acquisition demonstrated the most severe reconstruction artifacts. Random motion had the least significant effect on image quality, presumably due to the averaging effect of the motion over the duration of the acquisition. The 20% anterolateral defect was visualized in all cases, however, the apparent severity of the defect was altered considerably by the effects of cardiac motion.

Figure 4 compares the true motion of the cardiac phantom with that estimated by the four motion correction techniques for linear motion on either a single- or dual-head 90° gamma camera system. The CC technique failed to detect linear motion of the heart in the single-head study and only detected the large apparent shift in myocardial position between frame 15 and 16 in the dual-head 90° study. The DS, two-dimensional and MS methods all tracked the linear creep of the heart. However, the DS method significantly overestimated the magnitude of cardiac motion in the latter half of the study. The two-dimensional and MS methods accurately corrected for cardiac motion in both the single-head (Fig. 4A) and dual-head study (Fig. 4B).

Figure 5 presents the results for a single sinusoidal bounce of the heart during the acquisition, on either a single- or dual-head 90° gamma camera system. The motion detected by the CC technique was only poorly correlated with true cardiac motion. As with linear motion, the DS method overcompensated for motion in the latter half of the study, on both the single-head (Fig. 5A) and dual-head study (Fig. 5B). Both the two-dimensional and MS techniques accurately tracked motion of the heart. It can be seen from Figure 5, that all four techniques detected apparent motion in parts of the acquisition where no motion was present.

Figure 6A presents the results for continuous sinusoidal motion of the heart during the acquisition, on a dual-head 90° gamma camera system. The CC technique failed to detect any motion of the heart. The DS method accurately tracked cardiac motion over the first 90° of the acquisition but overcompensated for motion in the latter half of the study. Both the two-dimensional and MS techniques accurately tracked motion of the heart.

Figure 6B presents results obtained for random motion of the heart during acquisition, on a dual-head 90° gamma camera system. The CC method more accurately tracked cardiac motion in this acquisition than in the preceding studies, indicating that this algorithm is probably best suited to detection and correction of abrupt changes in cardiac position. The DS technique again accurately tracked cardiac motion over the first 90° of the acquisition, but overcompensated for motion in the latter half of the study. Both the two-dimensional and MS methods tended to slightly overestimate and underestimate, respectively, the true position of the heart.

Quantitative analysis of the average absolute error in determining the location of the heart is presented in Table 2 and shows that the MS and two-dimensional techniques yielded the best results for the six different acquisitions. With the MS and two-dimensional techniques, the average absolute error in each image in determining the true position of the heart was 1.5 and 1.7 mm, respectively, for the six studies. Both the DS and CC techniques demonstrated significantly higher average errors of 5.7 mm and 8.9 mm, respectively, and were inferior to the MS and two-dimensional techniques in all cases (p < 0.001). The implementations of the CC and DS technique used in this study were not optimized for analysis of images from dual-head 90° systems. Based on the findings that the DS technique incorrectly localized cardiac position in the latter half of the acquisition, we re-evaluated its ability to localize cardiac position in the four dual-head 90° studies based on analysis of the first 15 frames and applied these results to the last 15 frames. Results are shown in parenthesis in Table 2 adjacent to the original results. For the four dual-head studies, the average absolute error with the DS technique was 2.6 mm, compared with an error of 1.9 mm for the same four studies with the two-dimensional and MS techniques.

DISCUSSION

The adverse effects of patient motion on the quality of tomographic images of the heart have been well documented (9-11). Several studies have investigated the effects of axial and lateral motion and the timing of that motion on the quality of the reconstructed images (9,11,12). While it was not practical or feasible for such studies to simulate all possible combinations of axial and lateral motions, results generally showed that motion of two or more pixels (i.e., >13 mm) was required to create minor/moderate defects in the tomographic data. Only one study has shown that motion of ≤1 pixel (≤6.4 mm) can induce significant artifact (12). Two studies have investigated the timing of patient motion and have shown that motion occurring in midacquisition created more severe artifacts than those occurring at the beginning or end of the acquisition (9,10). Based on the above results, the magnitude of motion artifact in this study was set at ~2 cm (~3 pixels) and, where appropriate, to occur approximately midway through the acquisition. Acquisitions were also performed using either a single-head or both heads of a dual-detector 90° system, as a recent study by Cullom et al. (13) indicated that a 180° dual-detector 90° acquisition was significantly less sensitive to a given
FIGURE 3. Short-axis slices through same region of mid-ventricle for all seven acquisitions. A 20% defect had been placed in anterolateral wall of myocardium. No motion correction has been applied.

FIGURE 4. Correlation between true position of myocardium and that estimated by four motion correction techniques for (A) linear creep on a single-head system (acquisition 2) and (B) linear creep on dual-head 90° system (acquisition 3).

FIGURE 5. Correlation between true position of myocardium and that estimated by four motion correction techniques for (A) single sinusoidal bounce on single-head system (acquisition 4) and (B) single sinusoidal bounce on dual-head 90° system (acquisition 5).
motion than a corresponding single-detector acquisition. No attempt was made in this study to determine the magnitude of motion necessary for the creation of artifacts in the reconstructed images, as the main objective was to determine the degree to which different motion correction algorithms could adequately correct for patient motion in tomographic studies of the heart.

Attempts at eliminating patient motion artifacts in tomographic imaging of the heart have taken several different approaches. Germano et al. (1) adopted a proactive approach with the development of a multiorbit acquisition technique, which is particularly effective in eliminating artifacts resulting from myocardial creep. Other investigators have shown that increasing patient comfort through the use of prone imaging or support devices for the arms and lower back significantly reduces both the incidence and severity of patient motion (14,15). An intermediate technique developed by Germano et al. (2) has used an external point source positioned on the patient to facilitate postacquisition motion correction. Leslie et al. (3) showed that tracking movement of an external marker is significantly more accurate than tracking movement of the heart, although such external markers cannot correct for myocardial creep. Most work has centered around the development and evaluation of postacquisition processing techniques to correct for motion artifact (5–7).

In evaluating motion correction algorithms, previous studies have either performed postacquisition shifting of the individual images to simulate patient motion (5,6) or manually shifted the imaging table a discrete amount during the acquisition (3,7). Generally, motion has been simulated by a discrete shift of 0.5–3.0 pixels in the position of an image or a group of images in the acquisition. None of these methods accurately simulate many of the more common types of motion artifacts that occur in clinical studies, such as single/multiple bounces or “upward creep” (9). In this study, we have attempted to obtain a more realistic simulation of patient motion through the use of a computer-controlled sliding platform that permits an accurate simulation of various types of patient motion in the axial direction. This has permitted us to simulate motion of < 0.7 mm between images and incorporates the effects of blurring that would accompany motion in clinical studies. In addition, we were able to evaluate the effectiveness of these correction algorithms in dual-detector 90° studies. This is important because such systems are becoming the de facto standard for tomographic imaging of the heart.

Figure 4 shows that the CC technique failed to detect myocardial creep in both the single-head and dual-head simulations. This is due to the 1.8-mm threshold for motion detection with this technique, which is greater than the 1.3-mm motion between projections that occurs with the dual-head simulation of myocardial creep. This technique also failed to detect multiple sinusoidal motion of the heart (Fig. 6). In general, the CC technique performed poorly in all six simulations, with an average absolute error of 8.9 mm in determining cardiac position (Table 2). Previous studies using the CC technique have shown the technique to be capable of accurately correcting for sudden changes in the vertical position of the heart (5). This is partially confirmed in our findings with the CC technique shown both in Figures 4B and 6B. However, this technique was unable to detect the more subtle motion artifacts

![Graph](image-url)

**FIGURE 6.** Correlation between true position of myocardium and that estimated by four motion correction techniques for (A) continuous sinusoidal motion (acquisition 6) and (B) random motion (acquisition 7) on dual-head 90° system.

**TABLE 2**

Average Absolute Error in Centimeters Between True Location of Myocardium and that Estimated by Motion Correction Techniques

<table>
<thead>
<tr>
<th>Type of motion</th>
<th>No. motion correction</th>
<th>Cross correlation</th>
<th>Diverging squares</th>
<th>Two-dimensional ft</th>
<th>Manual shift</th>
</tr>
</thead>
<tbody>
<tr>
<td>Linear motion (single head)</td>
<td>0.98</td>
<td>0.98</td>
<td>0.58</td>
<td>0.13</td>
<td>0.10</td>
</tr>
<tr>
<td>Linear motion (dual head)</td>
<td>0.95</td>
<td>1.85</td>
<td>0.65 (0.16)</td>
<td>0.10</td>
<td>0.08</td>
</tr>
<tr>
<td>Bounce (single head)</td>
<td>0.26</td>
<td>0.31</td>
<td>0.49</td>
<td>0.14</td>
<td>0.08</td>
</tr>
<tr>
<td>Bounce (dual head)</td>
<td>0.52</td>
<td>0.88</td>
<td>0.57 (0.36)</td>
<td>0.20</td>
<td>0.17</td>
</tr>
<tr>
<td>Sinusoidal motion (dual head)</td>
<td>0.92</td>
<td>0.92</td>
<td>0.59 (0.22)</td>
<td>0.23</td>
<td>0.19</td>
</tr>
<tr>
<td>Random motion (dual head)</td>
<td>0.42</td>
<td>0.42</td>
<td>0.55 (0.32)</td>
<td>0.23</td>
<td>0.31</td>
</tr>
</tbody>
</table>

Average error 0.68 0.89 0.57 (0.26) 0.17 0.15

*Values in parentheses under diverging squares technique have been optimized for dual-head acquisition.*
seen in this study. This is in contrast to the findings of Leslie et al. (3), where the CC technique proved to be significantly better than the DS technique. A possible explanation for these conflicting results is the fact that previous studies have evaluated the CC technique with $^{201}$TI (3,5), whereas, in this study, we have simulated $^{99m}$Tc-sestamibi. The presence of simulated hepatic activity in this study may adversely affect the ability of the algorithm to detect motion. In addition, Leslie et al. (3) used background subtraction with the CC technique, which may have improved their results. In this light, the findings of this study and Leslie et al. (3) are not in conflict and merely indicate that the CC technique performs optimally in the absence of significant background activity.

In all simulations, the DS technique worked well over the first 90° of the acquisition, but consistently overestimated the true position of the heart in the latter half of each study (Figs. 4–6). The DS algorithm attempts to locate regions of high uptake relative to the surrounding background, hence adjacent or overlapping hepatic activity, particularly in the left posterior oblique projections, may cause the DS algorithm to incorrectly determine the true center of activity (7). The current implementation of the DS algorithm was not optimized for dual-detector 90° systems. We examined the effectiveness of the DS algorithm, using only the first 15 of 30 frames in the dual-head 90° acquisitions, thereby simulating implementation of this algorithm on a dual-detector 90° system. Table 2 shows that, under these conditions, the DS technique was significantly more effective in tracking cardiac motion and was only slightly poorer in performance than the two-dimensional and MS techniques. Overall, in this study, the DS technique appears to be superior to the CC technique (Table 2). Again, while these results are in conflict with previous studies by Leslie et al. (3) and Cooper et al. (10), these differences may reflect the use of $^{201}$TI in those studies compared with $^{99m}$Tc in this study. In addition, these discordant findings may reflect the type and magnitude of motion artifacts evaluated in these studies. In both previous studies, abrupt changes in cardiac location were introduced either by moving the imaging table or by shifting the image data. As discussed earlier, the CC technique is well suited to the detection of this type of motion, whereas the DS technique appears to be better suited to the detection of gradual cardiac motion.

By comparison with the CC and DS techniques, both the two-dimensional and MS techniques were more accurate in correcting for cardiac motion (Table 2). Results for the two-dimensional method are in agreement with those of Cooper et al. (10), who found the two-dimensional method to be the most accurate in measuring the distance of cardiac motion. Previous studies have not examined the effectiveness of a simple correction technique based on a manual shift of the images. The results from this study show that the human eye is a better judge of the true location of the heart than current automated techniques. The principal disadvantage of a manual technique is the time-consuming aspect of its application and the potential variability in results, depending on the experience of the operator. In our laboratory, a viable solution to this problem has been to first use an automated correction technique and then, if necessary, apply the manual technique to eliminate any residual motion not corrected by the automated technique.

In this study, simulating myocardial creep, there were incremental shifts in myocardial position of only 0.7 mm/image in the single-head study and 1.3 mm/image in the dual-head study. These small shifts between frames may pose a problem for many motion correction techniques and, indeed, are below the limits of detection for the CC technique as implemented on our system. Eisner et al. (12) have shown that noise in the image data can result in errors of up to 0.5 pixel (~3 mm) in the determination of cardiac position. Consistent with this, Germano et al. (2) have indicated that, even with an external point source, correction should only be applied for motion exceeding approximately 4 mm. Recognizing this potential limitation of many current motion correction algorithms, care should be taken when applying such algorithms to ensure that the algorithm does not introduce additional motion artifact into the clinical study, and that it appropriately detects and corrects linear creep.

As with most phantom studies, there are several limitations in how well our phantom model mimics real life. We have used organ activities that simulate those seen, on average, in a $^{99m}$Tc-sestamibi study (8). Clearly, the performance of the motion correction techniques will be altered in patients with significantly greater or lesser activity in the liver and bowel. This study did not simulate the effects of different radiopharmaceuticals, and it is possible that techniques that did not work well in this phantom model may work well in $^{201}$TI studies or in cases where there is minimal hepatic activity. Myocardial creep is not truly simulation in this model, since noncardiac activity also moves. This may affect the ability of some techniques, such as DS, to accurately detect such motion. This study was performed on a detector system with a relatively small axial field of view (25 cm). Larger-field-of-view systems will include more abdominal activity that will likely further degrade the ability of the CC technique to correct for patient motion. Finally, this study is limited to the simulation of axial motion, and it does not assess the effectiveness of motion correction techniques in compensating for lateral motion of the heart. However, previous studies have shown that lateral motion accounts for less than 3% of patient motion (9) and is generally less harmful to image quality than axial motion (10).

While it is obviously desirable to prevent patient motion, realistically, there will always be a significant percentage of patients in whom some motion will be present. Furthermore, with the increasing use of dual-detector 90° systems will come an increased perception of patient motion and an increased need for reliable motion correction techniques. We believe that an optimum strategy for addressing this problem should contain several elements: (a) use of prone imaging or support devices to increase patient comfort; (b) use of dual- or triple-detector systems to reduce acquisition times and, thereby, reducing likelihood of motion due to fatigue/discomfort; and (c) use of a combination of automated/manual correction techniques to remove any motion that does occur. Ideally, the user should have several different automated motion-correction techniques available to increase the likelihood of correcting for motion in all problematic studies.

**CONCLUSION**

Four motion correction techniques were evaluated using a cardiac phantom that realistically simulated many common types of patient motion in $^{99m}$Tc-sestamibi studies. While some of the correction techniques were limited in their ability to handle all types of motion, the two-dimensional fit technique and a simple MS technique yielded the most accurate estimations of the magnitude of cardiac motion and permitted artifact-free reconstruction of the simulated patient data.

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FIRST IMPRESSIONS

Technetium-99m-Methylene Disphosphonate (MDP) Bone Scan Hypercalcemia

PURPOSE

A 59-year-old woman was admitted to our hospital complaining of fatigue and somnolence. She was also suffering from weight loss, lumbosacral pain and temporary loss of orientation. Initial laboratory evaluation demonstrated elevated serum calcium (14.6 mg/dL) and lowered serum parathyroid hormone levels (1.8 pg/mL). Radiological examinations of the abdomen and thorax were unremarkable. Although the initial bone marrow biopsy findings were not satisfactory regarding the diagnosis of multiple myeloma, the patient was referred for a bone scan to detect eventual lytic bone lesions. Significant tracer uptake was observed in lungs, myocardium, stomach and biliary tract (Fig. 1). Lytic bone lesions were observed on posterior cranial views (Fig. 2). A definitive diagnosis underlying the hypercalcemia was not made; the patient died of renal and cardiac insufficiency.

TRACER

Technetium-99m-MDP, 740 MBq

ROUTE OF ADMINISTRATION

Intravenous

IMAGING TIME AFTER INJECTION

3 hr

INSTRUMENTATION

General Electric STARCAM 4000-I (GE Medical Systems, Milwaukee, WI) gamma camera

CONTRIBUTORS

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(Continued from page 5A)