Quantitative myocardial perfusion single-photon emission computed tomographic imaging: Quo vadis? (Where do we go from here?)

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Quantitative myocardial perfusion single-photon emission computed tomography can be improved further by technical advancements that are imminent in the clinical setting. These improvements are directed toward two main goals: (1) increasing the accuracy that the myocardial count distribution from tomographic slices represents the true tracer concentration and (2) increasing the accuracy of extracting this myocardial count distribution for quantitative analysis. Once these advancements are fully validated and implemented clinically, the clinical value of these cardiac diagnostic tests will be enhanced by increased accuracy of detecting and characterizing myocardial hypoperfusion and coronary artery disease. (J NUCL CARDIOL 1994;1:83-93.)

Key Words: single-photon emission computed tomography · myocardial count distribution · myocardial hypoperfusion · coronary artery disease

Judging by the large number of patients undergoing imaging with single-photon emission computed tomography (SPECT) to assess myocardial perfusion, it is clear that this technique has successfully attained widespread clinical acceptance. This is an opportune time to assess how to continue to improve this technique to overcome its present limitations and to ensure its continued growth and success. Quantitative analysis of these images is used clinically to objectify and standardize the process of image interpretation and should be considered a logical extension of the overall technique. Any improvement in image quality will directly influence the accuracy of the quantitative analysis. Thus both the imaging and quantitative analysis aspects need to be addressed when considering improving the overall technique. This article describes the technical advancements that are imminent in the clinical setting, as well as their expected clinical impact. These advancements should significantly improve quantitative myocardial SPECT imaging.

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DETECTOR/EQUIPMENT TECHNICAL ADVANCEMENTS

Multiple Detector Systems. For 180-degree tomographic acquisition of the heart, the optimal detector configuration is two heads separated 90 degrees apart. This is because the full 180-degree orbit may be acquired in half the time with only a 90-degree motion (Figure 1, *A*). There are four commercial vendors that currently offer such a device. For optimal 360-degree acquisition the triple-headed systems are preferred with each head separated by 120 degrees (Figure 1, *B*). There are at least four commercial vendors that offer these triple-headed devices. An important consideration is what to do with the gain in counts obtained from the additional detector(s). At Emory University we use a dual-headed, 90-degree separated detector system to increase throughput by reducing the total acquisition time by half and thus generating the same image quality but in half the time. If the choice is made to improve image quality rather than throughput, the consideration is to either increase the number of counts per image or switch to a higher resolution collimator and trade some of those counts to increase the spatial resolution. Although nuclear medicine techniques are inherently count poor, most of the acquisition protocols developed have already taken this into account. In general, trading the additional counts gained by the additional detector(s) to in-

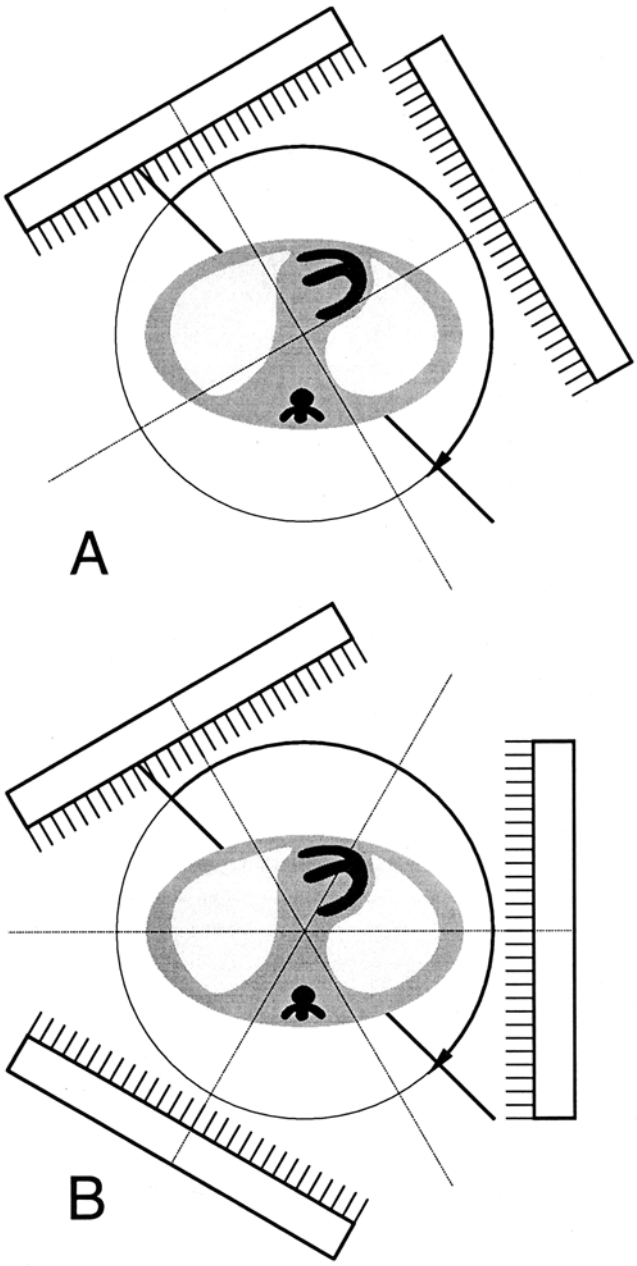

Figure 1. Dual versus triple-headed SPECT systems. **A,** Dual-headed system with detectors separated by 90 degrees can acquire 180-degree orbit by rotating through 90-degree angle, thus being able to image patient in half time compared with single-detector system. **B,** Triple-headed system with detectors separated by 120 degrees can acquire 360-degree orbit by rotating through 120-degree angle, thus being able to image patient in third of time compared with single-detector system.

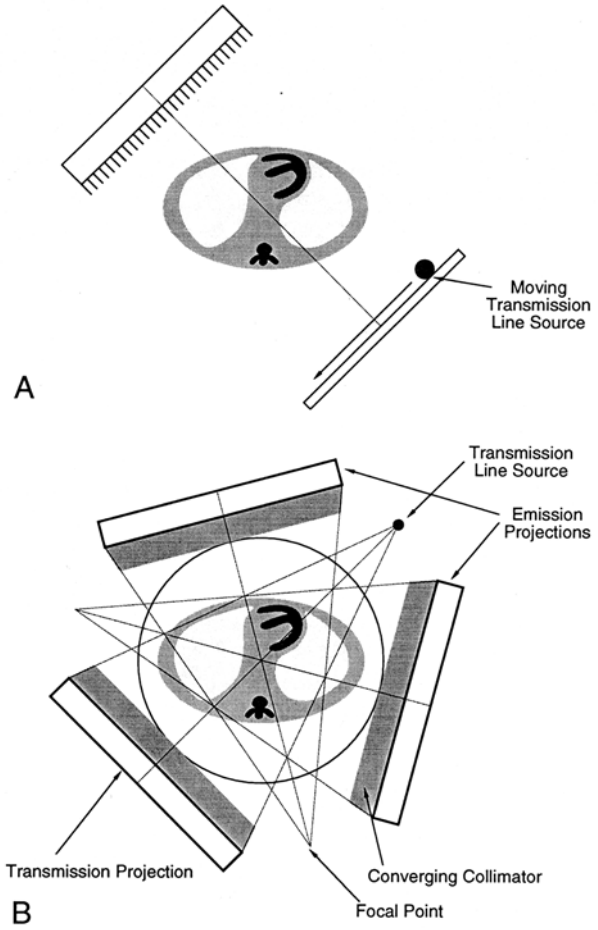

Figure 2. Two approaches for simultaneously acquiring emission and transmission projections. **A,** Radioactive (transmission) line source with energy different from that of tracer in heart is scanned across each projection. **B,** Transmitted photons from radioactive line source positioned at focal point of fan beam collimator are used to generate transmission scans, whereas two other detectors are used to acquire emission study.

crease spatial resolution results in a more meaningful improvement of both image quality and quantification. This finding, which was reported by Madsen et al.[1] for brain SPECT, also applies in myocardial perfusion SPECT. For example, in a ^{99m}Tc sestamibi study that uses a 1-day protocol the resting dose is 8 mCi, and the stress dose is about 22 to 25 mCi. This represents a difference in counts of at least a factor of 3, yet comparison of the image quality between the stress and resting studies shows that, except for very heavy patients, there are only minor differences.

Transmission Devices for Attenuation Correction. Before a SPECT myocardial study can be corrected for photon attenuation, a transmission study should be performed to account for the variable attenuation of photons traversing the thorax. There are several approaches that may be implemented clinically. Ideally, a system should be set up so that the emission and transmission studies may be acquired

simultaneously. One such approach has been suggested by Bailey et al.[2] In this approach a collimated radioactive line source (with energy different from the energy of the emission tracer source) extends across the field moving across the patient, scanning the patient once during each projection (Figure 2, *A*). The acquisition is electronically gated so that the system knows the exact location of this moving radioactive line so that the counts obtained at that location may be assigned as transmission counts once the counts emitted from the patient have been subtracted. At least one commercial vendor offers this device. Several companies are in the design stages of similar devices.

Another approach has been suggested by Gullberg et al.[3] This approach uses a multiple-headed SPECT system, in particular a triple-headed system. Two of the detectors, equipped with fan beam (converging) collimators, are used to acquire the emission information, whereas the third detector, also equipped with a fan beam collimator, is used to acquire simultaneously the photons transmitted across the patient originating from a radioactive line source located opposite this fan beam detector (Figure 2, *B*).

IMAGE RESTORATION TECHNIQUES

To realize the full clinical impact of these transmission scans, several image-restoration techniques that are already available need to be implemented. These are corrections for Compton scatter and photon attenuation. The spatial resolution of the imaging system is both limited and variable across the three-dimensional tracer distribution. This also causes a number of artifacts that may be corrected. Other important artifacts that need to be corrected or prevented are those associated with patient motion and heart motion.

Compton Scatter and Attenuation Correction Techniques. Galt et al.[4] and other investigators have implemented methods applicable to myocardial perfusion studies to correct for photon attenuation and scatter across the thorax. In these approaches planar projections of the myocardium are acquired corresponding to emission, scatter, and transmission images. The emission images are acquired with counts obtained from an energy window centered on the perfusion tracer's photopeak. The scatter images are acquired with counts obtained from a second, independent energy window placed over the Compton plateau portion of the energy spectrum, which provides an indication of the scatter contribution on the emission images. Once acquired, a fraction of the counts from the scatter images is subtracted from the

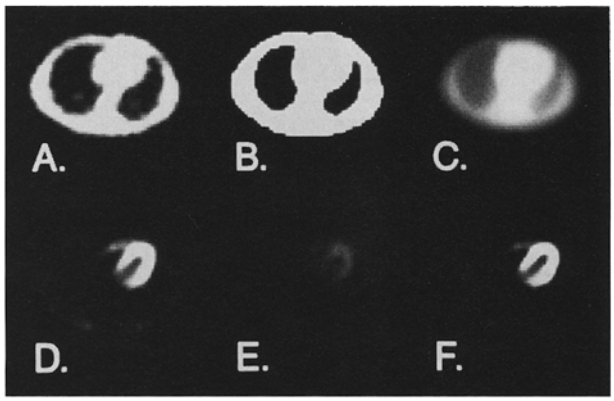

Figure 3. ^{99m}Tc sestamibi patient study, attenuation, and scatter corrections. *A,* Transmission transaxial of patient's chest. *B,* Map of assigned attenuation coefficients. *C,* Attenuation correction map. *D,* Corresponding photopeak emission transaxial slice through patient's myocardial walls and chambers. *E,* Scatter emission transaxial slice (scaled to show relationship to D. *F,* Scatter- and attenuation-corrected emission transaxial tomogram. (Used with permission. From Galt JR, Cullom SJ, Garcia EV. J Nucl Med 1992;33:2232-7.)

emission images to yield scatter-corrected images. The counts from the transmission study are used to reconstruct a low-quality computed tomographic image of the patient's thorax. From these transmission computed tomographic slices, a correction matrix is generated that, when multiplied times the scatter-subtracted transverse axial slices, yields scatter and attenuation-corrected images. When this simple technique is applied to a normal patient study, a better definition of the right and left ventricular myocardium is seen. This correction also yields more uniformity between counts at the base and apex of the heart and a higher contrast in the chamber of the heart (Figure 3).

More recently, Ye et al.[5] have shown that iterative algorithms may be applied to correct for Compton scatter, photon attenuation, and change of resolution with depth to yield a higher contrast in perfusion defects and a more uniform distribution of counts in normal myocardium from apex to base. In this three-dimensional iterative approach, the activity is reprojected back from each tomogram to each projection. Instead of just being projected back along a straight line, as is commonly done in filtered back-projection, the counts are dispersed along a cone of back-projection. This three-dimensional back-projection exhibits a more accurate representation of the imaging process. This and similar iterative techniques[6] are close to being ready for clinical implementation. The main drawback is that these tech-

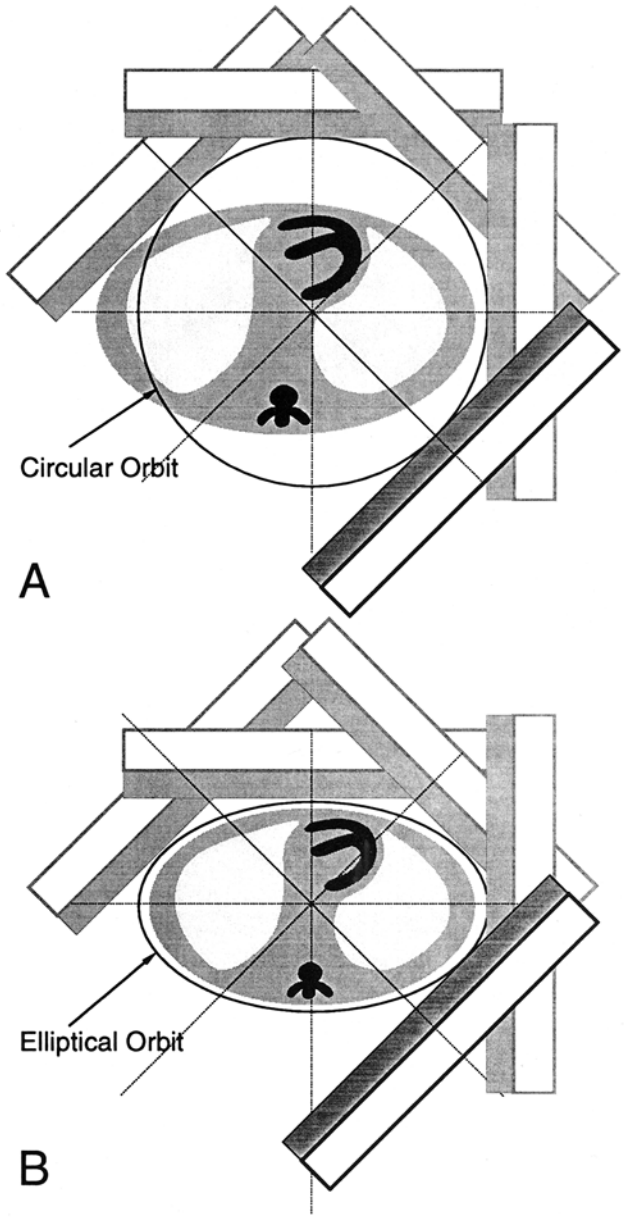

Figure 4. Elliptical versus circular orbits. **A,** One hundred eighty–degree circular orbit around thorax. Note that detector is closer to apex and anterior wall in 45-degree left anterior oblique projection compared with 45-degree right anterior oblique or left posterior oblique projections. Also note that 180-degree orbit is not centered on center of patient, thus allowing detector to be closer to heart compared with 360-degree orbit where patient's body has to be cleared on all sides. **B,** One hundred eighty–degree elliptical (or body contour) orbit around thorax. Note how much closer detector is to heart compared with circular orbit.

niques take additional time for the reconstruction process and are too time-consuming to be implemented in conventional (particularly out of date) nuclear medicine computers. Clinical implementation of these iterative reconstruction/correction algorithms is waiting for algorithm optimization to reduce the execution time. With the power of the superminicomputers already being used in our field, the optimization to allow these algorithms to execute in clinically acceptable times is imminent. There are numerous other approaches being developed by scientists in industry and universities to correct for Compton scatter, photon attenuation, and spatial resolution effects applicable to myocardial perfusion imaging. The main work that remains to be done before these techniques are widely accepted is to perform prospective multicenter trials that prove that their application would significantly increase the accuracy of detecting myocardial hypoperfusion by eliminating the physicians' need for reading around breast and diaphragmatic attenuation artifacts. Many of these techniques may be used to compensate for the differences in imaging with different energies, such as those of ^{201}Tl and ^{99m}Tc, thus also facilitating in the future the ability to perform and interpret dual-isotope imaging.

Relationship Between Orbits and Imaging Artifacts Caused by Spatial Resolution Effects. There have been many discussions during the years as to which orbit yields the best results. These arguments are centered around whether to use a 180-degree orbit that yields a higher spatial and contrast resolution or a 360-degree orbit that yields better field uniformity. Similar arguments are made for imaging with an elliptical orbit compared with a circular orbit. The key to accurate imaging with any of these orbits is to take into account, and correct for, the spatial resolution effects caused by the principle of collimation that occasionally creates imaging artifacts.

These effects of spatial resolution are best explained by comparing the field of view seen by the scintillation camera's crystal (NaI) through a single collimator hole of an all-purpose collimator compared with a high-resolution collimator. Each of these two collimator holes "sees" a conical field of view with the tip of the cone located in the collimator hole. In this configuration the angle of uncertainty described by these cones is wider for the all-purpose collimator compared with the high-resolution collimator. Unfortunately, when filtered back-projection is applied, counts that originated from anywhere within the conical region and passed through the collimator hole are back-projected along a single straight line perpendicular to the hole. Thus the assumption that the back-projection process places the counts back to where they originated can be wrong. Particularly, the farther the radioactive distribution being imaged is away from the detector the more incorrect this assumption is and thus the worst the resolution. This is another important reason for using the highest

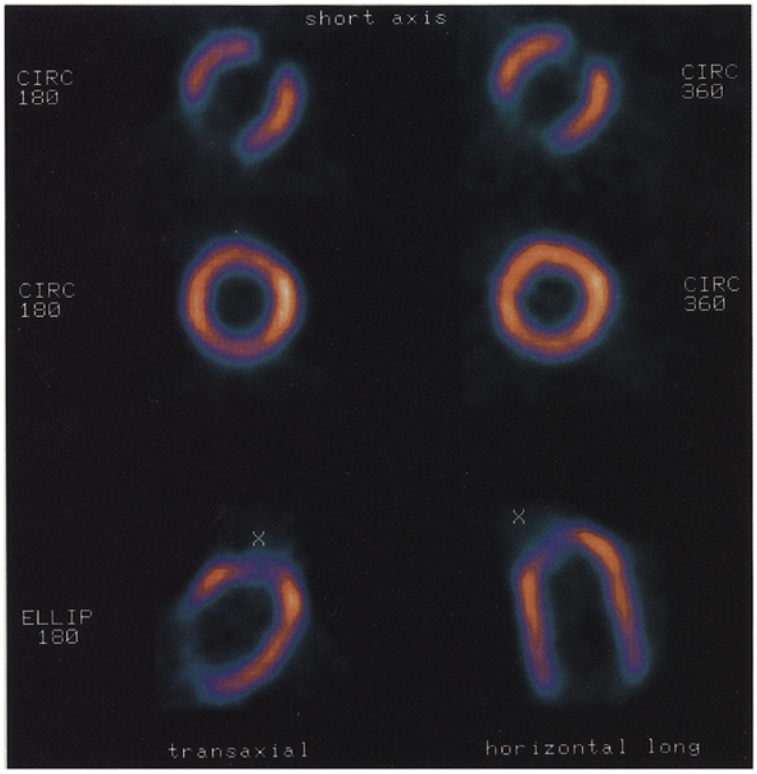

Figure 5. Comparison of defect contrast and normal slice uniformity in cardiac phantom. Comparison of defect contrast in two defects *(top panels)* and normal wall uniformity *(middle panels)* when imaging with circular *(CIRC)* 180-degree orbit and circular 360-degree orbit. Note higher defect contrast with 180-degree orbit and better uniformity (particularly anterior wall) with 360-degree orbit. Note photopenic artifact *(x)* generated with elliptical *(ELLIP)* 180-degree orbit.

resolution collimator that the existing counts will allow, particularly if accurate quantification is desired. If the study also has not been corrected for attenuation and scatter, the problem is compounded by the fact that the farther away a source is inside the body the greater the attenuation and scatter leading to more distortions that degrade quantification. By accurate quantification it is meant that the counts that are extracted from the patient's organs, in this case the heart, are directly proportional to the tracer concentration.

This unaccounted degradation of resolution with depth explained above often leads to imaging artifacts, particularly for 180-degree and elliptical orbits.[7] For instance, when acquiring with a 180-degree orbit, at some points in the orbit the detector is very close to the apex of the heart and at other projections the detector is far away from the same apical region. Thus the detector "sees" the same region on different projections with very different resolutions (Figure 4). Thus accurate quantification of studies acquired with a 180-degree orbit requires that these resolution effects be corrected. Although important, these effects are less important when acquired with a 360-degree orbit (Figure 5). This is because in this type of acquisition there is always an opposing view to the view that is close to the apex that is far away from the apex, which tends to average out the differences in resolution.

This disparity of resolution in the different projections is also manifested in artifacts when acquired with elliptical and circular orbits. This phenomenon may be seen in a phantom study. Short-axis slices of a phantom filled with uniform concentration of ^{99m}Tc and acquired with a 180-degree elliptical orbit result in a photopenic region in the apex because of this disparity in resolution (Figure 5). Thus even though acquisition with a 180-degree elliptical orbit would yield the highest spatial and contrast resolution possible, very few centers use it because of this artifact. Most centers are using a circular 180-degree orbit. Use of this orbit to image ^{99m}Tc myocardial distributions yields a reduction in count in the anterior wall compared with the septum or the lateral wall. This is because, even with a circular orbit, the detector does come close to the heart's anterior wall in some of the projections and farther away in others (Figure 4), so disparity of

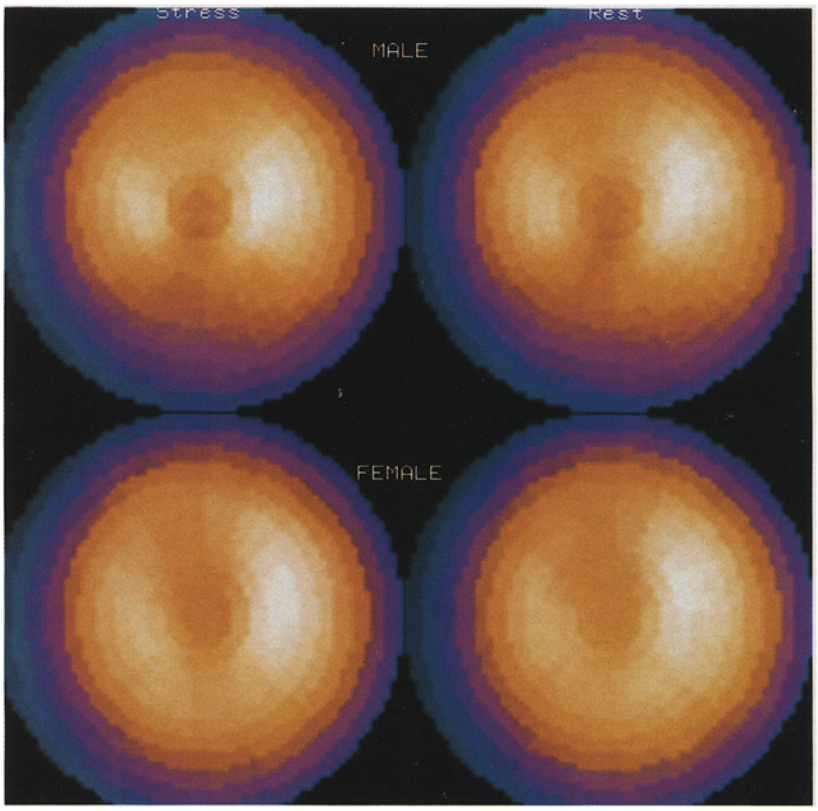

Figure 6. Polar maps of normal myocardial perfusion distributions of ^{99m}Tc sestamibi in patients with less than 5% likelihood of coronary artery disease. Note reduction in counts in anterior wall (11 o'clock) compared with lateral wall (3 o'clock) and distal septum (9 o'clock). Note differences in inferior wall (6 o'clock) distribution between male and female patients.

resolution can still exist, which creates artifacts (Figure 5). This reduction in counts is not seen when acquiring the same phantom with a 360-degree orbit (Figure 5). This same finding may be seen clinically by analyzing the mean normal tracer response of sestamibi in patients with a low likelihood of coronary artery disease acquired with a 180-degree circular orbit. A polar map displaying the mean normal response in these patients exhibits a photopenic groove in the anterior wall (11 o'clock), accompanied by an increase in counts in the septum and the lateral wall (Figure 6). Had the different correction methods already available been applied, a more uniform tracer distribution would have been expected, making it significantly easier for physicians to interpret these studies without having to read around these artifacts.

The important message of this section is that accurate imaging and quantification may be realized with either a 180- or a 360-degree elliptical or circular orbit, if these different resolution effects are taken into account.

Patient and Heart Motion. Accurate quantification requires that patient motion be prevented or corrected and the heart motion be frozen. Electrocardiographic synchronized gated SPECT applied to myocardial perfusion studies can be used to stop the motion of the heart. At Emory University all *stress* ^{99m}Tc myocardial perfusion sestamibi studies are acquired with this technique. Even though the tracer was injected at peak *stress* and is distributed according to myocardial perfusion, the patient is imaged starting at least 30 minutes after injection. Thus the gated acquisition is done at *rest* and the functional information obtained reflects *resting* function. Therefore in addition to stopping the heart motion by using gated SPECT, images are created that simultaneously provide the stress perfusion distribution and the resting functional distribution.[8] This attribute may soon obviate the need in many patients to acquire a separate resting perfusion study. Also, even though these end-diastolic or end-systolic gated images contain one fourth to one eighth the number of counts of the ungated images, they are both more quantitative and of higher quality than the ungated images because the smearing effect of the motion of the wall has been eliminated, yielding images of higher spatial

resolution (Figure 7). Without gating, there is a smearing of the counts coming from the tracer distribution in the moving walls, creating problems in interpreting the myocardial perfusion distribution. This is because if the same patient has two normally perfused walls, one that moves significantly and the other that does not, the counts from the wall that moves will be reduced because of this smearing effect compared with the other wall, even though they have the same tracer concentration. As described above there is also a direct dependence on the thickness of a wall and the counts that are recorded from that wall, as well as a dependence between the degree of thickening and the counts recorded on an ungated SPECT tomogram. As pointed out by Schmarkey et al.,[9] there is a significant effect that contractile dysfunction exerts on the ungated SPECT myocardial count ("perfusion") distribution. This effect has profound significance when interpreting the change in defect size or severity between a stress and resting study and when imaging "stunned" or "hibernating" myocardium.[9] Thus the use of gated SPECT should improve the accuracy with which the count distribution within the image represents the true myocardial perfusion distribution, assuming that there is an adequate number of counts in these gated images. The clinical impact of this improvement is still pending the application of this technique in large patient populations with a variety of cardiac disease states.

In terms of patient motion, many algorithms have been suggested to detect and correct for vertical motion along the axis of the table. Their success varies from patient to patient. The best approach is to try to prevent motion. The main reason patients are uncomfortable and tend to move is because their arms are tired from being stretched for more than 20 minutes of acquisition. There exists well-designed commercially available devices that provide support for the entire length of the bent arms at an angle that is natural for the patients to hold, eliminating the patients' discomfort in their arms and significantly reducing patient motion. Of course the use of multiple detector systems that reduce the total acquisition time also reduces patient discomfort and movement.

AUTOMATION OF PROCESSING STEPS

Other developments are addressing the need to reduce the number of operator-dependent processing steps. In the CEqual program, newly developed in collaboration between Emory University and Cedars-Sinai Medical Center to quantify the ^{99m}Tc sestamibi myocardial perfusion distribution, a number of operator-dependent processing steps have been automated. In this approach, Ezekiel et al.[10] have implemented algorithms for automatically finding the apical and basal slices, the center of the myocardium, and the maximum radius of search. These algorithms significantly reduce operator subjectivity and variability.

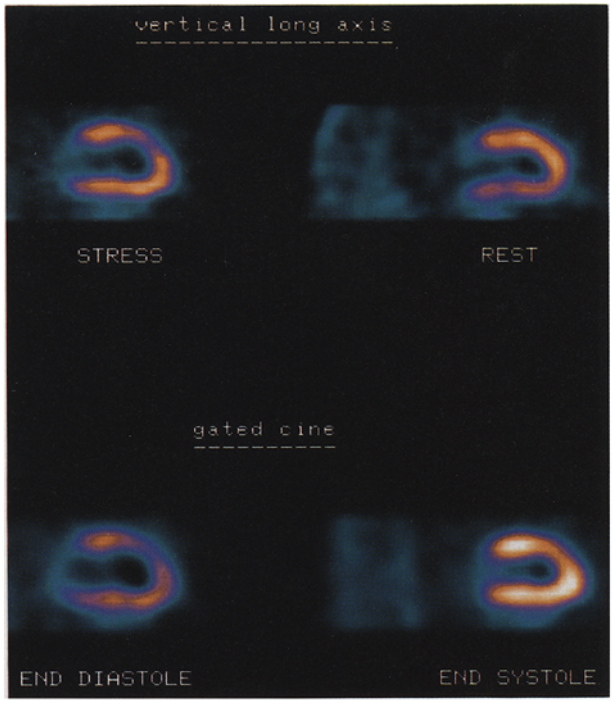

Figure 7. Comparison of ^{99m}Tc sestamibi ungated *(top panels)* versus gated *(bottom panels)* vertical long-axis slices. Note anteroapical perfusion defect in ungated stress study, which normalizes during resting study. Note how same defect in end-diastolic image becomes brighter (thickens) during end systole, predicting viable segment. Also note higher image quality of gated stress end-diastolic image compared with ungated stress image.

Another area of operator interaction associated with significant variability and subjectivity is the identification of the oblique angles of the myocardium to reconstruct the oblique axis of the heart. Once this step is automated, the entire procedure of reconstruction, quantification, and visualization will be performed without operator interaction and thus no subjectivity and significantly less variability. Mullick et al.[11] are pursuing a novel three-dimensional approach for detecting the axis of the left ventricular myocardium. The approach consists of automatically defining the endocardial and epicardial surfaces in terms of tiles. A vector perpendicular to each tile is then defined and the general direction it points to is used as the axis of the left ventricle. This axis is not necessarily a straight line. By using the principal

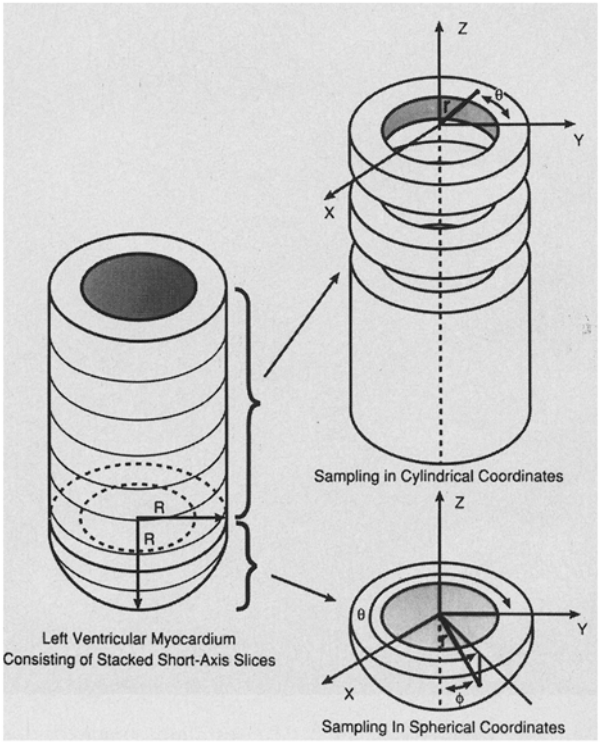

Figure 8. Three-dimensional hybrid sampling scheme. Scheme attempts to sample myocardial wall as perpendicular as possible to eliminate sampling problems. (Used with permission. From Garcia EV, Cooke CD, Van Train K, et al. Am J Nucl Cardiol 1990;66:23E-31E.)

components of this curve, an excellent correlation is obtained between the computer determinations and those of human experts at defining the axis of the heart. Future work will address the use of a curved rather than a straight axis of the heart and its implications on clinical interpretations.

QUANTIFICATION AND VISUALIZATION TECHNIQUES

Quantification of Myocardial Perfusion. Significant development has already been done in quantifying and using data-based approaches in three-dimensional myocardial perfusion distribution. As stated above, the latest development has been the approach used by the CEqual program to quantify ^{99m}Tc sestamibi distribution, which can also be applied to any other myocardial distribution. The main feature of this approach is the use of three-dimensional sampling by first stacking all short-axis slices and then using cylindrical coordinates to sample the basal portion of the myocardium and spherical coordinates to sample the apical portion[12] (Figure 8). This type of hybrid sampling ensures that the counts that are extracted are along lines perpendicular to the myocardial wall, thus avoiding sampling errors. Another feature of this approach is that, when each sample is extracted, the computer records its three-dimensional Cartesian coordinates. This three-dimensional information is then used by the computer to perform all quantifications in three dimensions and also to generate three-dimensional surface mappings of the myocardial tracer distribution. These algorithms are already commercially available through several vendors.

Quantification of Myocardial Wall Thickening. Even though the spatial resolution in SPECT is too low to measure the distance geometrically between endocardium and epicardium to yield myocardial thickness, Galt et al.[13] have shown that the change in myocardial counts throughout the cardiac cycle is proportional to the change in myocardial thickness. Thus if a region of the myocardium is followed throughout the cardiac cycle and the brightness in that region increases from end diastole (when the heart is dilated and the walls are thin) to end systole (when the myocardium thickens), it means that the wall is thickening and therefore it is viable. Therefore the change in counts throughout the cardiac cycle may be used as another marker of myocardial viability. Thus in addition to the fact that gated acquisition is necessary for either SPECT or positron emission tomography to freeze the motion of the heart to extract the accurate myocardial tracer quantification, there is also a significant amount of clinically relevant functional information that may be obtained from these studies.

Quantification of myocardial function from gated perfusion SPECT is being pursued according to the count-based approach stated above, which assumes that the change in counts extracted from the same region of the left ventricular myocardium is proportional to the change in myocardial wall thickness. Building on the concepts of Hoffman et al.,[14] Galt et al.[13] have shown that the counts extracted along the center of a cardiac phantom of uniform tracer concentration and varying wall thickness (Figure 9) will vary directly proportional to the thickness of the wall (Figure 10) up to a thickness that is about twice the spatial resolution of the imaging system ($\sim$26 mm). This relationship between myocardial thickness and extracted counts is being quantified by Cooke et al.[15] by using the first harmonic of the fast-Fourier transform to measure the amplitude and phase of the time-activity curve generated by extracting the counts along the cardiac cycle from each of the three-dimensional myocardial samples. Twice the ampli-

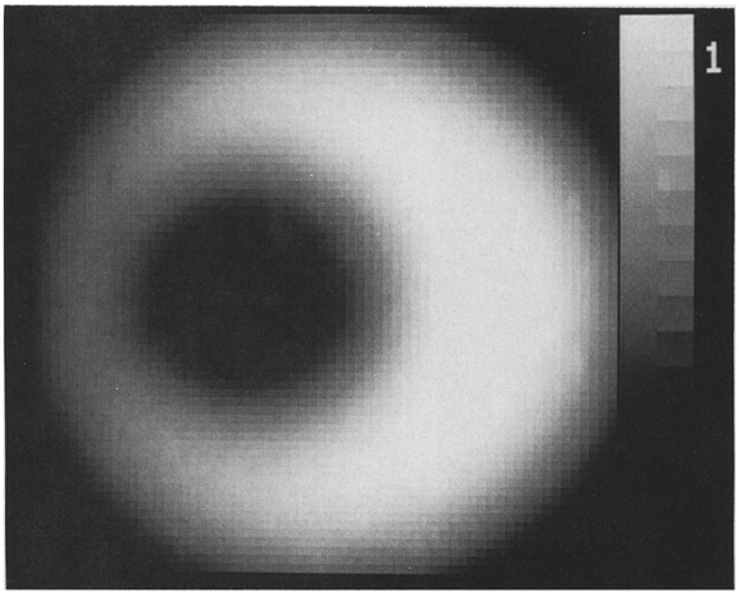

Figure 9. Simulated image of short-axis slice from cylindrical heart phantom with increasing wall thickness from left to right filled with uniform tracer concentration. Note how myocardium appears more intense in regions with thicker walls even though concentration is constant. (Used with permission. From Galt JR, Garcia EV, Robbins W. Effects of myocardial wall thickness on SPECT quantification. IEEE Trans Med Imaging 1990;9:144-50. ©1990 IEEE.)

tude of this curve yields the difference in counts between the end-systolic sample and the end-diastolic sample. Percent systolic wall thickening is then measured by dividing twice the amplitude by the end-diastolic counts multiplied times 100%. The phase information, obtained as the offset of one time-activity curve in one region of the left ventricular myocardium compared with another, is indicative of the onset of the wave of contraction. The three-dimensional mean normal response expected when measuring percent systolic wall thickening and phase has already been generated and applied in selected patient groups. Algorithm optimization for speed and extensive clinical validation is planned to be finished within 1 year. Completion of these tasks is required before this technique is ready for clinical use.

Image Visualization Techniques. Polar maps have been used for many years to portray the three-dimensional myocardial tracer distribution. Realistically, although many physicians are comfortable with this display, the true three-dimensional left ventricular myocardium does not resemble a disk like the one used in polar map representations. The techniques exist now to make the transition to allow visualization of this information in a true three-dimensional display. It is impossible with a two-dimensional polar map to represent the three-dimensional myocardial tracer distribution to accurately

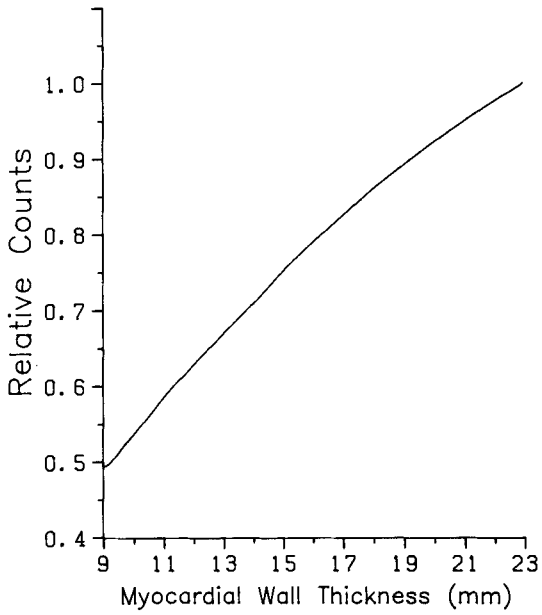

Figure 10. Dependence of maximum counts versus myocardial wall thickness of cylindrical heart phantom described in Figure 9. Note how there is near-linear relationship between myocardial wall thickness and maximum counts extracted. (Used with permission. From Galt JR, Garcia EV, Robbins W. Effects of myocardial wall thickness on SPECT quantification. IEEE Trans Med Imaging 1990;9:144-50. ©1990 IEEE.)

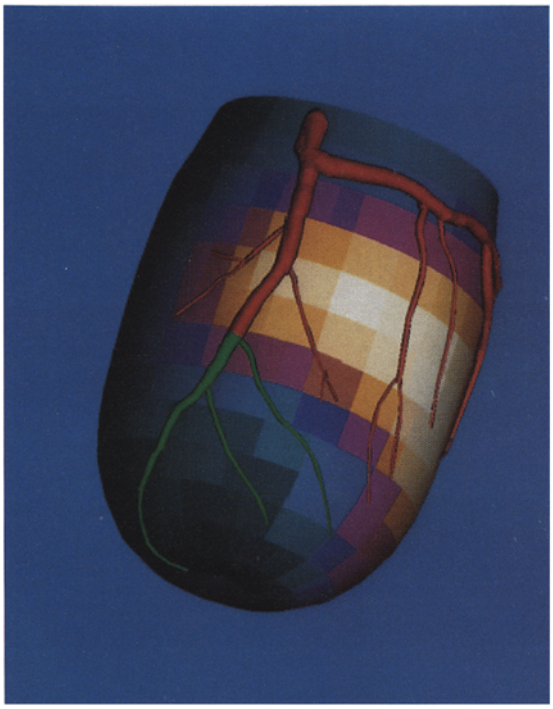

Figure 11. Registration of coronary tree obtained by use of angiography onto ^{99m}Tc sestamibi myocardial perfusion distribution. Transition from red to green portion of artery indicates where a balloon occlusion of left anterior descending artery occurred. Darker colors in surface map represent regions of hypoperfusion. Note excellent agreement between site of occlusion and extent of perfusion defect.

portray both the true extent and location of the abnormality. Although newly modified volume-weighted and distance-weighted polar maps can yield accurate defect extent and location, respectively,[12] the only way to accurately display both the true extent and location of the defect in the same display is to use a three-dimensional approach.

The three-dimensional display is also the ideal platform for unifying information from various imaging modalities. In a collaborative project between Emory University and the Georgia Institute of Technology, the three-dimensional myocardial perfusion distribution acquired with SPECT is being fused with the three-dimensional rendering of the coronary arterial tree obtained from biplane digital angiography and reconstructed by novel mathematic techniques[16] (Figure 11). A computer algorithm then uses the anterior and inferior interventricular grooves from the perfusion study as landmarks to align the left and right coronary vasculature, respectively. This approach attempts to unify the physiologic information extracted from the myocardial perfusion study with the anatomic information extracted from the coronary angiographic study to provide both complementary and conflicting information. It is expected that the use of these unified displays in patients in whom the anatomic and physiologic findings are decoupled (what appears as conflicting results) will yield a better understanding of coronary artery disease pathophysiology. Clinical validation of this unification process is about 3 years away.

Decision Support Systems. There are many tools already available from the field of artificial intelligence that may be applied to myocardial perfusion imaging. One such useful tool is expert systems. In this approach the knowledge of experts may be captured in terms of rules used by the experts to interpret myocardial perfusion scans. In another collaborative effort between Emory University and Georgia Tech University an expert system (PERFEX) has been developed to use the output of the polar map quantitative programs to suggest which walls are hypoperfused, which of the hypoperfused walls are reversible (associated with myocardial ischemia) and which are fixed, which defects are associated with imaging or processing artifacts, and which vascular territories are jeopardized.[17] All of these conclusions are provided with a certainty factor that indicates how certain the program is of each conclusion. The system also allows the physician to ask for justification of each of the conclusions. The program then provides a list of the rules, stated in English, that were used to reach the final conclusions and how each rule changed the level of confidence of the outcome. Programs such as this will assist the physicians being trained, or those with less expertise, to interpret studies at a consistently high level of expertise. These programs still require extensive validation before they are disseminated because of the potential for their improper use. This is expected to be completed in the next 2 years.

SUMMARY

The direction of these technical advances dictate the direction that the clinical use of myocardial perfusion SPECT will be taking if physicians, scientists, and technologists continue to work together. These directions are to continue to use multiple detectors, to use transmission sources for attenuation correction, to start applying attenuation correction algorithms clinically including application of scatter correction algorithms, to use higher resolution collimators including correction of the spatial resolution effects, to use multiple-gated SPECT to allow quantification of myocardial function, as well as to obtain

a more accurate perfusion distribution, to correct for the differences in scatter distribution in dual-isotope imaging, and to use three-dimensional displays of myocardial tracer concentration and decision support systems.

Once these advances are fully validated and implemented clinically, the clinical value of these cardiac diagnostic tests will be enhanced by increasing patient throughput, improving image accuracy, increasing the utility of the test, reducing interobserver and interoperator variability, increasing the clinical accuracy of the test, and, ultimately, reducing the cost of the test.

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