ponent measured in our ^{99m}Tc-teboroxime patient population. However, it should be considered that motion-induced artifacts in ^{99m}Tc SPECT studies are probably more severe than analog artifacts in ^{201}Tl SPECT studies for the same amount of motion and the same pre-processing/reconstruction filter cutoff, given the higher resolution capabilities of technetium-based agents. Conversely, we have not performed a quantitative or qualitative assessment of the changes in sensitivity and specificity for ^{99m}Tc-sestamibi or ^{99m}Tc-teboroxime myocardial SPECT studies following the application of our motion correction method. Future extensions of this preliminary work will include validation of the technique in a prospective patient population using quantitative analysis to clearly assess the clinical significance of this motion correction strategy.

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EDITORIAL

Effect of Motion on Cardiac SPECT Imaging

There are several major theoretical problems with cardiac SPECT imaging: attenuation, scattering, changes in biodistribution during acquisition, changes in resolution with depth, nonuniformity and nonlinearity of the detector(s), errors in the center-of-rotation, and so forth. Two articles in this issue discuss another problem: motion during acquisition. There are two categories of motion during SPECT acquisition. The whole patient can translate or rotate with respect to the camera or the heart or surrounding organs can move with respect to the rest of the body. An example of the second type of motion is "upward creep" of the heart after exercise, which is probably caused by changes in respiration (*1*). It is truly remarkable to me that with all of these theoretical problems, cardiac SPECT imaging has become a useful clinical tool.

The process of tracking an object in an image is a frequent image processing operation used in tasks as disparate as Landsat imagery, cruise missile navigation and radiologic image registration. The methods in the papers by Germano et al. (*2*) and Cooper et al. (*3*) in this issue represent two general approaches: tracking a fiducial mark or tracking a feature in the image. Fiducial marks can be designed so that they can be accurately and reliably tracked, however, as in myocardial imaging, it is not always possible to affix a fiducial marker to the object of interest.

Germano et al. use a point source on the sternum as a fiducial marker to track and correct for whole-body motion. Cooper et al. track the image of the heart, which tracks both motion of

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the patient and motion of the heart with respect to the rest of the patient. Although tracking the heart can correct for both sources of motion, it assumes that the heart is of similar configuration in sequential projection images.

The first step in identifying and tracking features in sequential images is often to preprocess the image in order to increase the energy of the feature with respect to the ground. For example, an interesting problem arises in satellite imagery when comparing a feature with changed vegetation or with snow cover. The next step is to select a feature to be tracked, for example, the myocardium. A similarity measure is used to reflect the similarity between the feature and a specific portion of the second image. The similarity measure is defined as a function of the offset between the feature and the search area in the second image. A frequently chosen similarity measure is the cross-correlation of the feature and the search region. If the feature being tracked is not the highest energy object in the image, then the cross-correlation is typically normalized for the energy in the window being examined (*4*). Using interpolation, the feature can often be located with subpixel accuracy, often on the order of one-tenth of a pixel accuracy. (Accuracy depends upon image distortion and the autocorrelation function of the feature).

In the paper by Cooper et al., three different similarity measures were compared to visual interpretation of the cine display. One method used cross-correlation of the projections of successive frames unnormalized for energy (*5*). A second method, diverging squares, tried to find a square with the maximum intensity (*6*). This square was assumed to be located around the myocardium. The third method used the sum of the squares of the difference between images. Normalized two-dimensional cross-correlation was not used. Each of these methods worked reasonably well. However, none of the methods would have accurately measured all motions which could affect the reconstructed images.

It is conceivable that other algorithms might track the heart even more accurately. However, the heart changes in sequential images due to changes in the viewing angle, overlying attenuation and background activity. Thus, even a much more complicated algorithm which took into account much of this *a prior* information might not perform any better than the tested algorithms.

As would be expected, Germano et al. were able to track a point source on the sternum with much greater accuracy. However, their algorithm takes into account neither rotational motion nor motion of the heart with respect to the sternum. These authors argue that most motion-related reconstruction artifacts are due to motion of the whole patient. Certainly many of the worst reconstruction artifacts are caused by motion of the whole patient. Thus, tracking only whole patient motion may provide the information necessary to correct almost all of the important motion related reconstruction artifacts.

If we step back a moment, there are a number of important points to be made. First, a general rule is always to collect the best possible raw data. Attention to technical detail must also be stressed. Data correction schemes can be used to ameliorate poor data but it is always best not to have to correct for bad data. Second, it is essential to control quality for data collection problems. The physician interpreting the data should view the raw data cine and should know what corrections have been used. Third, the physician needs to understand how reconstruction artifacts are produced and needs to know how to recognize artifacts in his system (*7*).

The decision about how to handle patient motion may also involve the other data collection problems listed in the first paragraph. Kiat et al. have reported that there is reduced motion artifact in the raw data using prone imaging (*8*). Prone imaging probably also reduces "upward creep" of the heart. Prone imaging has a major impact on decreasing inferior wall attenuation artifacts, but may result in some increase in apical and anterior wall artifacts. We favor prone imaging, but many departments still prefer supine imaging.

The decision about prone versus supine imaging and about patient motion correction will have to be revisited when accurate attenuation correction becomes clinically available. The papers by Germano et al. and Cooper et al. in this issue provide us with important data to help us make these decisions.

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