TEACHING EDITORIAL

Scatter Correction for SPECT

The problem of Compton scatter in single photon emission computed tomography (SPECT) has received relatively little attention in the literature. Unlike the more serious problems of photon attenuation and camera nonuniformity, which can produce dramatic artifacts in SPECT images, scatter causes only some loss of lesion contrast, slight blurring of the edges of organs, and some increase in apparent radioactivity. But in general its effects are subtle and not very disturbing. Nevertheless, as one tries to refine techniques and algorithms to achieve more accurate reconstruction of the true radioactivity distribution, one must eventually address the problem of Compton scatter.

There are at least four approaches to scatter correction. The first, which uses an "effective" linear attenuation coefficient (*1*), is already used (often unknowingly) in most instances in which some sort of attenuation correction is carried out. The true attenuation coefficient for the 140-keV photons of Tc-99m in water is $\mu = 0.15\ \text{cm}^{-1}$. If this value is used for attenuation correction in a reconstructed cross section of a large phantom filled with Tc-99m, the radioactivity in the phantom will not appear uniform as it should, but will "bulge" from the periphery to the center. The excess radioactivity represents the detected Compton scatter, whose reconstructed image is superimposed on the uniform image produced by the detected primary (unscattered) photons after attenuation correction. Now if one uses a smaller attenuation coefficient, say $\mu = 0.13\ \text{cm}^{-1}$ for brain images and $\mu = 0.12$ or smaller for abdominal images, then one can undercorrect for attenuation in such a manner that the "dip" due to the remaining attenuation effect just offsets the "bulge" due to the scatter, resulting in a uniform radioactivity distribution in the reconstructed image. This approach, in effect, replaces some of the primary photons lost through attenuation with scatter photons that are detected in their place. It is a crude form of scatter correction, since it does not address the problems of loss of contrast and edge sharpness caused by scatter.

The second approach involves estimation of the scatter component of the image through computer modelling (*2,3*). Simulations of the true radioactivity distribution and of the surrounding scattering medium are stored in the computer, and a scatter image is generated from these data either through mathematical techniques involving matrix manipulation (*2*) or through Monte Carlo techniques, in which the histories of a large number of scattered photons are simulated and traced to determine the probability of detection (*3*). This scatter image is subtracted from the actual detected image. The difference, representing the primary photon component, is then corrected for attenuation using the true value of μ ($0.15\ \text{cm}^{-1}$). In theory this approach should result in accurate scatter correction, but in practice it is limited by the "truthfulness" of the simulations, is quite difficult to implement, and makes very heavy demands on the computer.

The third approach, recently reported in the *Journal* by Axelsson, Msaki, and Israelsson (*4*), involves scatter correction of images by deconvolution. The assumption is made that the scatter component blurs the image of the primary photons in a constant and predictable manner. The nature of this blurring is determined by imaging a line source in a scattering medium. Axelsson et al. have demonstrated that the blurring functions for superficial and deep radioactivity differ appreciably, so an average blurring function is used. The scatter component is removed from images by deconvolving them with this function. This approach would be expected to work reasonably well in situations with relatively little superficial activity, as in renal imaging; however, one would expect that in situations with a prominent component of superficial activity, as in liver imaging, it would lead to an artifactual "overcorrection" for scatter.

The fourth approach, reported in this month's *Journal* in an article by Jaszczak, Greer, Floyd, Harris, and Coleman (*5*), attempts to measure the scatter component directly as the Tc-99m image is acquired. This is done by collecting a separate image in a scatter window (92 keV to 125 keV) at the same time that an image is being collected in the primary window (127 keV to 153 keV). The reasonable assumption is made that the events detected in the scatter window are related to the scatter component of the events detected in the photopeak window by a constant factor k. The authors found this factor to have a value of approximately 0.5 in a typical imaging situation, based on measurement and also on Monte Carlo simulation. Their approach then consists of reconstructing separate photopeak and scatter images of the radioactivity distribution; subtracting the scatter image, weighted by a factor of 0.5, from the photopeak image; and applying an attenuation correction to the resulting image using the value $\mu = 0.15\ \mathrm{cm}^{-1}$. They demonstrate that this approach yields an accurate correction for Compton scatter with improved lesion contrast and edge sharpness, and permits quantification of the radioactivity distribution.

What is the place of scatter correction in SPECT imaging? Since the effects of scatter are subtle, its removal cannot be expected to produce dramatic improvements in image quality. In fact, one probably should not begin to worry about scatter correction until one is convinced that the other sources of artifacts—namely, system misalignment, camera nonuniformity, photon attenuation, and patient motion—have been properly addressed. In the meantime, acceptable results can be obtained merely by using an appropriate "effective" attenuation coefficient in the attenuation-correction algorithm.

When one is ready to face the scatter problem, the approach of Jaszczak et al. appears to be a good solution. It does require a dual-energy detection capability, and it will double the computation time, but the data-acquisition time should not increase. The authors found the value of the factor k to be 0.5 for a 22-cm phantom, but its value for other phantom sizes remains to be determined, although the accuracy of the method is probably not strongly dependent on the precise value of this factor. While the method should produce sharper images with better lesion contrast, it will also increase image noise, and in fact the signal-to-noise ratio is not improved, as Jaszczak et al. point out. Nevertheless, some improvement in lesion detection might result if scatter correction shifts the image of the lesion into a portion of the gray scale where it becomes more obvious. The increased image noise might be handled by using a filter with a slightly lower cutoff in the reconstruction algorithm.

The main advantage of the approach to scatter correction proposed by Jaszczak et al. is that it brings us closer to the elusive goal of truly quantifiable SPECT. These authors have found that the SPECT count rate accurately measured the true radionuclide concentration for spheres greater than 2.5 cm in diameter in their scatter-compensated images. In contrast, the SPECT count rate for a 6-cm photon-deficient sphere overestimated the true radionuclide concentration by about 30% (relative to background) when the "effective" attenuation-coefficient approach was used. The ability to make accurate measurements of radioactivity from SPECT images will, we hope, lead to methods for quantification of total and regional organ blood flow and function far superior to methods currently available. This will enhance our ability as nuclear physicians to determine the extent of disease processes and the changes occurring over serial studies.

BERNARD E. OPPENHEIM
Indiana University School of Medicine
Indianapolis, Indiana

REFERENCES

1. LARSSON SA: *Gamma Camera Emission Tomography.* Acta Radiol Suppl 363. Stockholm, Acta Radiologica, 1980, p 32
2. EGBERT SD, MAY RS: An integral-transport method for Compton-scatter correction in emission computed tomography. *IEEE Trans Nucl Sci* NS-27:543–548, 1980
3. BECK JW, JASZCZAK RJ, COLEMAN RE, et al: Analysis of SPECT including scatter and attenuation using sophisticated Monte Carlo modeling methods. *IEEE Trans Nucl Sci* NS-29:506–511, 1982
4. AXELSSON B, MSAKI P, ISRAELSSON A: Subtraction of Compton-scattered photons in single-photon emission computerized tomography. *J Nucl Med* 25:490–494, 1984
5. JASZCZAK RJ, GREER KL, FLOYD CE, et al: Improved SPECT quantitation using compensation for scattered photons. *J Nucl Med* 25:893–900, 1984