EDITORIAL

Quantitative Emission Tomography

Single photon emission computed tomography (SPECT) with rotating scintillation cameras has been under development for over 10 years. During this evolution, one of the primary objectives has been to provide accurate quantitative cross sectional images. A similar pursuit has existed in positron emission tomography (PET) for even longer. The incentive is obvious: if activity per unit volume in vivo can be measured accurately, information from tomographic imaging will be enhanced creating opportunity for advancing investigational techniques into clinical applications. Not only will more accurate activity distributions improve qualitative image interpretation, but it will make it possible to quantitate rate constants of physiologic models, metabolic processes, and radionuclide kinetics for dosimetry. A decade has passed since the pursuit for quantitative SPECT imaging began, yet we are still not there. The difficulties that must be overcome are several, including variable spatial resolution with distance from the detector, limited energy resolution adding Compton scattered photons, statistical uncertainties in count density, and correction for attenuation of the photons between each voxel and the detector for each projection.

Corrections for attenuation are substantial and more complex with SPECT than with PET. For example, at 140 keV only ~20% of the photons are transmitted through 10 cm of tissue, whereas, for the 511 keV annihilation photons from positron emitters, ~40% are transmitted. PET has several advantages over SPECT for quantitative results. Present day PET cameras have higher intrinsic spatial resolution, thereby providing better delineation of the target volume. At these higher energies, there is also improved scatter rejection. Also, in a uniform medium, attenuation correction is easier because the probability of coincidence detection of the annihilation photons is independent of the positron location along the line of flight of the photons through the object, dependent only on the total distance along the line of flight. This is quite different than for single photon emission.

Mechanisms for attenuation correction in SPECT can be grouped into those that modify the projections before image reconstruction and those that reconstruct the image and then apply a correction to the image (*1*). Most assume or measure the body contour. For iterative reconstructions, the corrected emission distribution is used to reproject the data to compare with the measured profiles and modified to minimize the difference. Many of the prereconstruction corrections are based on the assumption of uniform attenuation through the object and modification of the projections by an average attenuation factor. These first order corrections work best for a uniform distribution of activity. The postreconstruction corrections modify the pixel values in the reconstructed image. The technique of Bailey et al. described in this issue of the Journal, uses the measured attenuation values at each pixel to correct the reconstructed emission distribution (*2*).

Their approach is simultaneous acquisition of emission and transmission data. The transmission data, using the lower energy photons of gadolinium-153 (^{153}Gd), are corrupted by scatter from the higher energy technetium-99m (^{99m}Tc) emission photons. They demonstrate a technique to uncouple these components by predicting both the number and the distribution of scattered photons in the lower energy window and then subtracting this from the measured distribution. Although their scatter model is not theoretically exact, the results suggest that it works quite well. They also demonstrate that the measured attenuation coefficients for ^{153}Gd can be used to predict those of ^{99m}Tc. There are several very attractive features of their technique if it proves generally valid following more extensive phantom and clinical tests. First it is a simultaneous acquisition. Sequential acquisitions of transmission and emission data have been previously reported (*3*). This has been particularly helpful in PET imaging because of the short half-lives of many of the positron agents. It is not generally applicable to SPECT unless the emission agent concentrates very rapidly so that there is little waiting time after injection and the patient does not have to be repositioned between the acquisitions. Even so, two acquisitions would be necessary. Other advantages listed by these authors are the availability of an anatomic image from the transmission data to complement

the emission images, which also provides individual body contours for each section for attenuation correction, and very low additional radiation absorbed dose. Their claim of 10 mrem additional dose is typically <10% of the dose to the target organs for most imaging procedures. The calculated attenuation values by this approach will be less than expected for 140 keV in tissue because of the inclusion of scatter using an uncollimated sheet source and a 20% window with a NaI(Tl) detector. This is similar geometry to the emission image and, therefore, a reasonable first order approximation for inclusion of both attenuation and scatter.

The problem of depth dependent spatial resolution is at least partially circumvented by using the geometric mean of opposed views. The spatial resolution and its variation over the images is of central importance in quantitating radioactivity concentration in the determination of the object volume. It has been shown previously that the accuracy of activity concentration measurement is a function of the object size with reference to the system resolution (*4*). Additionally, the statistical limitations in emission computed images are compounded when modified by noisy transmission data for attenuation correction. These authors have not addressed this aspect in this paper but evidently have minimized the problem, as seen by their excellent quantitative results with measured and known activity concentrations within 5%.

Attempts to correct for attenuation and the other factors that limit quantitative SPECT has been increasing in recent years as evidenced by the variety of techniques published in the literature and presented at the Society of Nuclear Medicine meetings. For example, the influence of depth dependent resolution and attenuation correction and the introduction of distortions by these variables is described in a recent paper by Eisner et al. (*5*). Also, a different approach to image reconstruction incorporating attentuation, resolution changes with depth, and scatter has been proposed by Floyd et al. in their inverse Monte Carlo protocol (*6*). These are but two recent examples emphasizing the effort going into the solution of the problems currently limiting quantitative radioactivity in vivo.

The technique of Bailey et al., for "Improved SPECT Using Simultaneous Emission and Transmission Tomography" appears to be a very convenient and practical approach to quantitative SPECT. We will look forward to additional phantom verification and patient application.

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