Comparing Diagnostic Accuracy of γ Camera Coincidence Systems and PET for Detection of Lung Lesions

TO THE EDITOR: A recent issue of *The Journal of Nuclear Medicine* published 2 articles on measuring the diagnostic accuracy of detecting lung lesions by coincidence detection. Both articles reported high detection accuracy for lung lesions with γ camera coincidence (GCC) systems that are comparable with PET. One article, by Tatsumi et al. (*1*), reported an elegant study that compared the diagnostic accuracy of detecting lung lesions by GCC with PET. A second article by Weber et al. (*2*) measured the diagnostic accuracy of detecting various sizes of lung lesions using GCC.

The physics of the GCC and PET systems would predict much better results with PET than with GCC because of the better performance characteristics of PET compared with GCC. However, the difference in detection accuracy reported in the 2 articles is not so great. Therefore, the question is why does the GCC system with poorer resolution, lower sensitivity, and no attenuation correction perform as well as the PET system for the detection of lung nodules?

Coleman et al. (*3*) have shown using the "Coleman lung phantom" that spheres of 6, 10, 13, and 22 mm in diameter, simulating lung nodules, were detected accurately by PET. However, GCC missed detecting the 6-mm-diameter sphere, and SPECT missed detecting the 6- and 10-mm-diameter spheres. They also found that not doing attenuation correction resulted in geometric distortion of the spheres and reduced contrast between sphere activity and background activity. In another publication, Weber et al. (*4*) showed, with clinical imaging of lung nodules and lymph nodes with FDG PET, that small lymph nodes are missed with GCC compared with PET. Several other studies confirm the finding of Coleman and Weber.

To understand better the results of the 2 studies by Tatsumi et al. and Weber et al., we need to address the roles of image resolution, lesion size, lesion contrast, and image noise in the measurement of diagnostic accuracy. An elegant study by Lim et al. (*5*) describes the relationship of these parameters to detectability of lesions with nuclear imaging systems and how they affect diagnostic accuracy. I shall summarize, in lay terms, the basic roles of resolution, lesion size, and image noise in the detectability of lesions.

The basic theory for detection of lesions can be summarized by stating that the detection of a lesion is proportional to the size and the contrast of the lesion and is inversely proportional to reconstructed resolution and the statistical noise in the image. The larger the lesion, the higher the detectability. The higher the contrast of the lesion to the background, the higher the detectability. The smaller the resolution, the better the detectability, and the lower the statistical noise in the image, the better the detectability of the lesion. High detection accuracy for lesions is easily achieved with large lesions that have high contrast. Given large lesions and high contrast, even a SPECT imaging system can detect lesions accurately as shown by Coleman et al.

Two different resolutions are used in PET: 1 specified by the physicists in describing the system characteristics (the intrinsic resolution) and another used during clinical imaging (the recon-structed or system resolution). No PET cameras reconstruct clinical images to the intrinsic resolution of the cameras. To do so would increase the image noise associated with inverse filtering in the reconstruction process to an unbearable level, and the images would be unreadable. So, most cameras reconstruct clinical images with reconstructed resolutions that are 20%–100% worse than the intrinsic resolution, depending on parameters such as the type of clinical application, scan times, dose injected, and counts collected. GCC systems typically reconstruct poorer resolution than do dedicated PET systems, and the physics of detection would predict lower sensitivity for GCC than dedicated PET for lung nodules.

Lesions that are larger than twice the resolution of the system are detected linearly with respect to the recovery of data from the lesion. Smaller lesions suffer a loss of contrast because of the partial volume effect of the small lesion. Several articles describe the physics of partial volume effects and loss of contrast in small lesions. There are also phantoms that can be used to measure the partial volume errors. But, stated simply, a lesion should be at least 2 times the reconstructed resolution in order not to suffer a loss in contrast. In general, large lesions are easier to detect than are small lesions, and the sensitivity for detecting large lesions will be similar for GCC and dedicated PET.

The greater the contrast between the lesion and the background, the easier it is to detect the lesion. Tumors with high metabolic rates are easily detected using FDG imaging compared with tumors with lower metabolic rates. With FDG PET, it is easier to detect lesions in the lungs than in the liver because of the lower background FDG uptake in the lungs compared with the liver. The low density of lung tissue contrasted with the higher density of tumor automatically produces a 3:1 contrast for tumors located in the lung. Therefore, lung lesions will typically have higher contrast and will be detected better than liver lesions.

All nuclear images have statistical noise associated with the limited number of counts detected. In general, for uniform distribution of radioactivity, the lower the number of counts in the image, the higher the statistical noise. The best way to reduce image noise is to increase the sensitivity of the PET camera, increase the scan time, or increase the injected dose. GCC systems have extremely low detection efficiency because of the thin detectors. Typical detection efficiency of the GCC system is <3% of a modern PET camera per axial centimeter of field of view. Therefore, image noise is much higher in GCC than in dedicated PET cameras, and GCC systems are not as good as dedicated PET systems in detecting small differences in contrast.

It is distressing to note that some PET and GCC users are advocating the use of no attenuation correction for the detection of tumors with FDG PET. One of the major advantages of PET is the ability to do attenuation correction and improve the uniformity of detection in the field of view for better detection of lesions. With attenuation correction, small nodules located centrally can be detected with PET (*4*). Not doing attenuation correction reduces the ability to detect these centrally located lymph nodes and makes the test for detection of metastasis less than optimum. Determining lymph node involvement is very important for the staging of cancer patients.

Sacrificing attenuation correction is not the answer in PET and GCC, especially since segmented attenuation correction can reduce

the total scan time for attenuation. We should not make the same mistake as with SPECT imaging of the heart, where the lack of attenuation correction has resulted in thousands of misdiagnosed and equivocal results.

An unwritten law in physics states that you don't get something for nothing. The current GCC systems with thin sodium iodide crystals are not as good from the physics of detection as are the dedicated PET systems. Therefore, they will not be able to detect small lesions as accurately as PET. Coleman et al. and Weber et al. showed with phantom and clinical studies that small nodules are missed by the current GCC systems but not by PET. Lymph nodes, which are small and are involved with lung metastasis in the mediastinum and the hilar regions, are as important as detecting solitary nodules in the lung. Detection of small lesions with high accuracy will require detection systems that are properly designed to do so. Comparing the diagnostic accuracy of GCC and PET for detecting lesions has to be done with a full understanding of the impact of resolution, lesion size, system resolution, and statistical noise in the images. Selection of these parameters in a clinical protocol can influence the outcome of the results, and this is why standard phantoms such as the Coleman lung phantom should be used to characterize the detection of lesions with different detection systems.

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Does Attenuation Correction Work?

TO THE EDITOR: The editorial by Dr. Wackers (*1*) and the article by Vidal et al. (*2*) in the August issue are timely. Having been involved for some time in developing methods for acquiring transmission data for subsequent use in attenuation correction (*3,4*), it has been both encouraging to see the widespread interest in producing correct reconstructions of the distribution of radiopharmaceutical and disappointing to see that the adoption of the methods has become so contentious. In early studies by our group in Sydney we discovered that correcting properly for attenuation was only part of the total solution to the problem of producing artifact-free reconstructions. In our first clinical report on the outcome of attenuation correction in ^{201}Tl myocardial perfusion scanning in 11 patients, with angiographic correlation in 7, we reported (*5*) the following:

"In 2 patients with normal right coronary arteries and no past history of inferior infarction, inferior wall defects erroneously identified using NC (no correction) were correctly reported as normal using AC (attenuation correction). In addition, one patient with a 90% LAD lesion showed an anterior wall defect only with AC. Thus, the use of AC led to 3 additional cases being correctly reported. Conversely, in one patient with a low likelihood of CAD but no coronary angiography, AC demonstrated an anterior defect whereas NC was normal. In the remaining 7 cases, there was no difference in final diagnosis between AC and NC."

As Dr. Wackers correctly points out, many other factors may influence the reconstruction of myocardial perfusion data. These include variable resolution with depth, choice of reconstruction algorithm, partial volume effects, patient movement, cardiac motion, respiratory motion, and scattered photons. There are several other possible sources of error in addition to these in the study of Vidal et al. They include the following: truncation of the emission and transmission data caused by the use of fanbeam collimators, leading to incomplete projection data and possible artifacts in the reconstructed data; lack of a downscatter correction from ^{201}Tl (167 keV) into the ^{99m}Tc window (140 keV), especially in the region of the heart where transmission is low and the emission counting rate is reasonably high; and lack of photopeak scatter correction in the ^{201}Tl (72 keV) window.

We have spent some time in our group in London examining one commercial version of the ^{153}Gd scanning line-source approach for simultaneous emission/transmission scanning (Vantage; ADAC Laboratories, Milpitas, CA). We found that downscatter from the 100-keV ^{153}Gd photons to the lower 72-keV ^{201}Tl window, one of our main concerns, was sufficiently low that it could be ignored. In spite of this, though, we found initially that image quality had been compromised because of a different factor, the slight decrease in counting rate caused by the electronic transmission window, and corrective action was required. This is exacerbated by the relatively low ^{201}Tl doses (74 MBq) permitted for myocardial perfusion scanning in the United Kingdom.

We propose that, on the basis of exhaustive testing that has been reported in the scientific literature, we accept that attenuation correction in heterogeneous data works (*1,2*). However, testing in the laboratory is often different from commercial realizations of a method. Therefore, we agree with Dr. Wackers that new, novel acquisition schemes should be proven to work, and we should not simply accept the manufacturer's word that their implementation of the system described by another group produces identical results. In this area, the proposal to have more standardized phantom testing should be endorsed. However, we do not believe that this is going to solve the problems that have been highlighted in the article by Vidal et al. and the editorial by Dr. Wackers.

If we accept the proposal that attenuation correction does work, we are then left with the question as to what is causing the confusing results in the articles cited, especially in the anterior wall and apex of the heart. It is certainly possible that correcting properly for attenuation will enhance some physical errors—for example, scatter will be worse toward the center of the body and will be increased preferentially relative to the edge of the body. Also, lack of scatter correction will artifactually redistribute reconstructed counts into areas of lower attenuation such as the lungs. Motion may play a role in the change in the reconstructed activity toward the apex of the heart between corrected and noncorrected data.

It seems to us that 1 way forward is to try to separate these effects and study them in isolation, if possible. Phantom testing has