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Design and Use of PET Tomographs: The Effect of Slice Spacing

TO THE EDITOR: Miller et al. (*1*) have elegantly explained and demonstrated the requirements of axial sampling in PET. Their findings are consistent with the work done by Mullani (*2,3*) with theoretical and phantom studies and with Senda et al. (*4*) with clinical PET studies except in their conclusions.

As a designer of PET cameras with considerable experience in three-dimensional imaging camera design (*5*), I have to disagree with Dr. Miller's conclusion, which is stated as follows. "Thus, for typical objects, the slice spacing of the tomograph should be approximately 50%–75% of the z-axis resolution (FWHM)." My reasons for disagreeing with this conclusion are discussed below.

1. The Nyquist sampling criteria for repetitive functions such as a sine function is to sample at twice the frequency of the function, which could be interpreted as slice separation divided by slice resolution (S/FWHM) ratio of 0.5 (50%). However, for non-repetitive functions such as a Gaussian function, accurate recovery of information requires the sampling frequency to be greater than twice the "resolution frequency," which is approximated by 1/FWHM. In the transaxial direction, most PET designers have found it necessary to sample between two to three times the "resolution frequency" (33%–50% ratio for sampling distance to resolution) in order to avoid spatial artifacts in the image due to aliasing. I have found similar spatial artifacts in the axial direction when imaging a special partial volume phantom (*3*), if the axial sampling is inadequate. Dr. Miller's conclusion that the sampling/resolution ratio can be reduced to 0.75 (75%) is much coarser than that specified by Nyquist sampling criteria and certainly not consistent with the sampling requirements in the transaxial direction.
2. Dr. Miller's simulations were carried out with highly symmetric and smooth objects—spheres. The frequency content of the sphere as contrasted to a bar of radioactivity across the field of view is quite different and the aliasing effects will be quite smaller for the spheres than for the bar of radioactivity. Undersampling a bar of radioactivity will certainly result in greater aliasing which in turn will result in greater underestimation of the recovery coefficient. Therefore, his estimate of 10% acceptable error in recovery coefficient by under sampling may be higher for objects such as the myocardium and the cortical areas of the brain which are closer in geometry to a bar of radioactivity than a sphere.
3. Acceptable error in recovery coefficient is arbitrary and will depend significantly on the clinical application. A 10%–15% error in quantitation may not be significant for a moving heart application where there is considerable smearing of radioactivity from one slice to another due to the motion of the heart. However, for gated heart images, where the motion is minimized, it could be interpreted as a mild defect since a 10%–15% change in a physiologic condition such as myocardial perfusion can be clinically significant. Similarly, in oncology where one might encounter spherical objects, a 10% error may not be acceptable in a serial study where a tumor is studied pre- and post-treatment to assess changes in metabolism or shape of the tumor. A 10% change in tumor quantitation may be clinically significant in the treatment strategy for the patient. It must also be remembered that undersampling causes spatial artifacts which will result in an error in the estimation of the size of the tumor. Again, for a serial study in which the size of the tumor is an important measure of the effect of treatment, an error of 10% may not be acceptable.

I have been a proponent of finer axial sampling in PET for several years and have deliberated a great deal about the compromise between the optimum and the acceptable number of slices and slice separation. I agree with Dr. Miller that finer sampling puts a great deal of burden on the finite number of photons collected in a PET study. Fewer number of photons per slice will result in poorer image quality and a greater uncertainty in quantitation within the slice. Finer axial sampling will decrease the number of photons per slice and the temptation to decrease the number of slices by decreasing the sampling distance between slices is very strong for a PET camera designer. However, for the best three-dimensional imaging, the sampling criteria for the transaxial and the axial directions must be satisfied in a similar fashion. It is not appropriate to arbitrarily apply a different sampling criteria in the axial direction than the transaxial direction. For that reason, the S/FWHM ratio cannot be higher than 0.5 (50%) and in a practical sense should not be much higher than 0.4 (40%) for accurate three-dimensional reconstructions in PET.

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REPLY: We appreciate the careful study of our paper by Mr. Mullani, an experienced and innovative PET researcher. We believe the points raised in his letter do not affect the conclusions drawn in our paper. Each of his points are addressed below.

1. The "resolution frequency" is an ill-defined concept for a blurring function, such as a Gaussian, which tapers off continuously. Thus, we chose not to use that concept in our paper. Instead, we computed the correct integrals exactly and left it to the reader to draw conclusions from the graphical data. The fact that our results are expressed in units of full-width at half maximum (FWHM) does not mean that this measurement was assigned any special significance. Data derived from Mullani's phantom do not refute our conclusions.

2. The frequency-domain characteristics of a bar and a sphere are indeed different. The case of a flat object, such as a myocardial wall lying in the transaxial plane, can be treated by analysis of a "slab" of activity with infinite extent in the transaxial plane and finite axial dimension. Thus, the integral expressions reduce to the simple one-dimensional form. Figure 1 shows a new analysis for this case presented along with the three-dimensional data from our paper. Note that the uncertainty in the recovery coefficient (E_{rc}) is actually less for the one-dimensional case than for the sphere, a result apparently opposite to that suggested by Mullani. This finding is intuitively plausible if one considers that the maximum activity in a sphere declines with increasing axial offset because of transaxial blurring of the progressively smaller cross-sectional area of the sphere. The infinite slab has constant activity as a function of offset except near the edge of the bar. Note that the recovery coefficient for a sphere will be lower than that for a bar of the same thickness; however, our analysis refers to the *variability* of the recovery coefficient with position, not its actual value. Results for the aliasing measure (Q_a) are also very similar for the slab and the sphere.

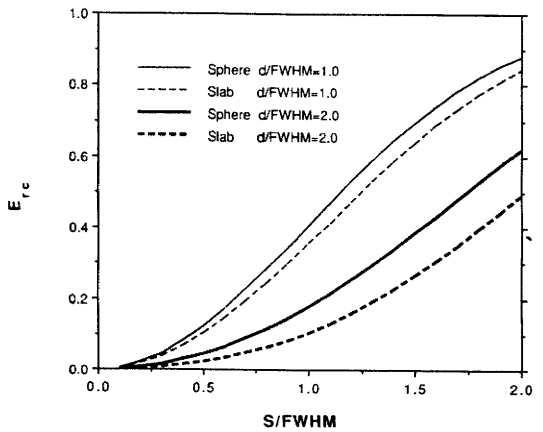

FIGURE 1
The measure of variation in the recovery coefficient, E_{rc}, is shown as a function of the ratio of slice spacing-to-resolution (S/FWHM). Results are shown for the two ratios of object size-to-resolution (d/FWHM) for a sphere and a slab.

3. We agree that the acceptable sampling error depends on the application. That is the reason we presented complete graphical data; the investigator or tomograph designer can choose the appropriate slice spacing according to the imaging situation. Note that variability in the recovery coefficient only falls to zero for infinitely close spacing.

In summary, we believe our principal conclusion, slice spacing should be approximately one-half the full-width at half-maximum, is valid. We believe we are in basic agreement with Mullani. In fact, his great practical experience with tomograph design strengthens our shared opinions.

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Geometric Methods for Determining Left Ventricular Volume

TO THE EDITOR: The article "Left Ventricular Volume Calculation Using a Count-Based Ratio Method Applied to Multigated Radionuclide Angiography" by Massardo et al. (*1*) adequately delineates the disadvantages of geometric methods for determining left ventricular volume by radiocardiography and describes the limitations of heretofore reported count-proportional nongeometric methods. The authors describe the theory and application of a "count-proportional reference volume" method for determination of left ventricular volumes. They imply that this method avoids the pitfalls of geometric methods and retains the advantages of a nongeometric count proportional technique without the need for blood sampling and attenuation correction.

I suggest that this implication is erroneous. The method described is nothing more than a geometric model employing a sphere and an indirect measurement of its diameter rather than the more sophisticated prolate ellipsoid as described by Dodge et al. (*2*) for contrast angiocardiography and as applied to radiocardiography by Strauss et al. (*3*).

Consider the prolate ellipsoid representing the left ventricle (LV) generated by rotation of the ellipse

$$\frac{X^2}{\left(\frac{L}{2}\right)^2} + \frac{Y^2}{\left(\frac{S}{2}\right)^2} = 1,$$

where L and S are the long- and short-axes, respectively.