Editorial: Advances in Emission Tomography: Quo Vadis?

This issue of *The Journal of Nuclear Medicine* contains three articles representing advances in emission tomography instrumentation (*1–3*). Common to all three designs is the goal of acquiring volumetric data with good sampling in three dimensions. There are two motivations leading to the volumetric data acquisition. First there is a physiologic and clinical need to take into account the complex nature of the organ such as the brain in *x*, *y*, and *z* directions. Secondly, there is an argument that improved sensitivity will come from systems whose detectors can acquire data from a larger solid angle than provided by positron emission tomography (PET) instrument design with single crystal detector units and septal shields between layers. This editorial is written to serve three purposes:

1. To highlight the distinguishing features of these three instrumentation papers.
2. To present quantitative comparisons between the single-photon emission computed tomography (SPECT) and PET instruments.
3. To forecast some ideas for the future developments in both SPECT and PET.

The two new PET instruments have achieved good transverse resolution of <6 mm and continuous transaxial sampling thus avoiding interleaving gaps which necessitate a second scan in many conventional scanners if full volumetric sampling is desired. These systems differ one from the other radically in almost every aspect of design from different scintillators and detector arrangements (see Table 1) to different reconstruction algorithms.

POSICAM 6.5 (*1*) achieves a continuous sequence of 21, 5.125 mm sections using an ingenious staggered detector design wherein every other crystal around a transverse section is axially offset one-half a crystal height. Modules of 11 crystals and 6 photomultiplier tubes (PMTs) are arranged like barrel slats around the imaging port and septa. This instrument has very good sensitivity.

The PENN-PET device (*2*) utilizes six Anger-logic cameras in a hexagon without employing septa to remove scatter. Operation without septal shielding increases the sensitivity but also increases the scatter fraction and degrades the spatial resolution at the object edge. PENN-PET scatter reduction is remarkably well handled by energy lower threshold adjustment such that it is possible to operate with only 20% scatter fraction using a 350-keV–400-keV threshold and no septa (Table 1). At first glance this result seems too good to be true when one considers that with bismuth germanate (BGO) an energy threshold of 350 keV and septa, the scatter fraction is 18%. The explanation lies in the fact that NaI(Tl) has eight times more light output than BGO and its energy resolution is so superior that a 350-keV–400-keV threshold does a much better job of excluding scattered events. But when one makes a comparison of the sensitivity between the POSICAM 6.5 and PENN-PET for the same scatter fraction, we note a difference by a factor of 2.5. A major factor which explains this difference is the greater efficiency of BGO crystals. However, if we used a 100-keV lower energy threshold for the PENN-PET, the measured sensitives (*2*) (after subtraction of 40% scattered events and randoms) would be comparable after adjusting for the 20% longer axial field of view (FOV) of POSICAM 6.5.

QUANTITATIVE COMPARISONS

The pertinent question regarding comparison of sensitivity is: How do we compare sensitivity under conditions where the scatter fraction is different from one machine to another? The problem here is that an effective sensitivity value must include an adjustment for the influence of the noise contributed by the scattered events (scattered coincidences), randoms (accidental coincidences), and the noise of the true unscattered coincidences. An effective method first used by Robert Beck in gamma (Anger) camera studies is to derive a figure of merit (*4*), which was adopted by Derenzo (*5*) and others (*6*) to optimization of PET ring detection shielding.

The effective sensitivity Q is given by:

$$Q = \frac{\text{Trues}^2}{\text{Trues} + \text{Scattered} + \text{Randoms}}.$$

Pertinent to the comparison of contemporary PET designs in this issue is a very recent study of Thompson (*7*) in which this figure of merit known as "noise-effective count-rate" is calculated for both individual crystal-phototube arrays and for block detector PET designs having either NaI(Tl) or BGO crystals. That extensive work, based on Monte Carlo simulations, gives quantitative results which bear careful study when evaluating the meaning of sensitivity figures based on the total number of detected events.

We learn from these simulations that there is about a five-fold increase in true coincident count efficiency when the inter-plane septa are removed from a PET system and all off-axis events are recorded. But there is

Received Feb. 22, 1990; revision accepted Mar. 1, 1990.
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TABLE 1
Design Differences in Three New Pet Systems

	POSICAM 6.5 PET	PENN-PET	Osaka SPECT
Crystals	1320 BGO	6 NaI(Tl)	4 NaI(Tl)
PMTs	720	180	120
Axial range of view (cm)	12.3	10	17
Width of view (cm)	58	50	22
Resolution			
Transverse (mm)	5.8	5.5	7
Axial (mm)	11.9	5.5	7
	12.8 (10 cm)	8.4–9.5 (10 cm)*	
Sensitivity			
cps/μCi/cc	148,000†	48,000‡	§
cps/μCi/cc/axial cm	12,000	4,800	814¶
Scatter fraction:			
% at 350–400 keV	18	20	
% at 400 keV		10	
Saturation μCi/cc in FOV	2.4	1.8	24.5**

* Improvements to <7 mm feasible with implementation of a known three-dimensional algorithm.
† Published value is 180 kcps from which 18% is subtracted.
‡ Value from Figures 7 and 8 for scatter of 20% (*2*).
§ Value given is 101 cpm/37 kBg or 1.7 cps/μCi. (point source per head).
¶ Estimate based on conversion of a 1.7-cps/μCi value given in (*3*) to a theoretical result for a 20-cm diameter phantom with theoretical scatter fraction of 0.6 (*8*).
** Estimated from sensitivity of 1.7 cps/μCi and saturation rate of 222 Kcps.

a more than two-fold increase in scatter fraction for a 15-cm long cylinder. Thus, the effective sensitivity is not the trues as measured by the recorded coincident events minus scattered and random events; true sensitivity is the product of the trues and the ratio of the trues to the recorded coincident events.

SPECT has a theoretical sensitivity 21 times less than the radionuclide sensitivity of PET for resolution of 7 mm (Fig. 1). Devices for SPECT which have approached this theoretical value do not have the volumetric coverage adequate to image the whole brain in one rotation. The major point of the system design presented by Osaka University and Hitachi Corporation scientists (*3*) is to achieve a volumetric coverage and high sensitivity such that studies can be made of tracers whose distribution might change in short periods of time. The innovation here is in the arrangement of the four sodium iodide rectangular crystals backed by 30 PMTs each in a rigid box configuration as close as possible to the patient's head. The use of compact arrays of 5 cm square PMTs allowed an important reduction of detector edge-to-field of view distance. From this arrangement one can expect a system sensitivity four times greater than that of a single-headed tomographic

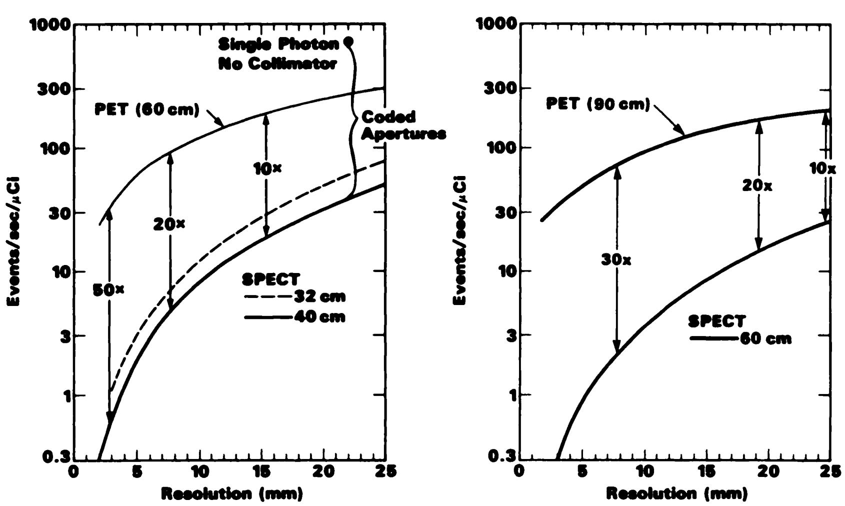

FIGURE 1
The relationship between sensitivity of SPECT and PET for both head and body instruments for a point source in a 20-cm diameter phantom for single layer systems. Multiply by 314 for cps/μCi/cc per axial centimenter. Efficiency (E), packing (f), and attenuation (α) for PET: $E^2 = 0.8$, $f = 1.0$, and $\alpha = 0.4$; for SPECT: $E^1 = 1.0$; $f = 0.6$, and $\alpha = 0.5$.

instrument, and ~1.3 times greater than the three-sided SPECT units currently available for whole-body studies. Unlike the larger three-sided systems, the Osaka four-headed system is optimized to the head for brain studies with care to provide fast and stable rotation options. Although the system design emphasizes rotation speeds of 360° in 10 sec, it was not clear that there is adequate sensitivity to acquire dynamic data with acceptable statistics even with 60-sec rotation speeds. The sensitivity for 7 mm resolution in Table 1 is more than 10 times less than the highest sensitivity quoted, which was for 21 mm resolution. From the numbers given in (*3*), The authors and I calculate an axial sensitivity using conventional PET units. Based on the assumptions made, the measured sensitivity is ~5.6 times less than that of the PENN-PET device and 14 times less than the POSICAM. This is close to the expected factors if one takes into account the differences in detector dimensions, crystal efficiencies, and resolution (Fig. 1).

Count-Rate Capability and System Saturation

A problem for all three systems in this issue is the count-rate capability. The deadtime as measured by the time required to measure the energy and position of each event will be smaller for a system with many detectors than one with a few large crystals or large blocks of detectors viewed by a lesser number of PMTs. This deadtime and detector pulse pileup can limit a system's ability to perform without saturation if 10's of millicurie amounts of short half-time tracers are used in dynamic studies. For PET systems, the highest count-rate capability is expected for systems that have the maximum number of parallel channels. This is achieved when each detector crystal is coupled to a single PMT, however, this direct-coupling design might not give the optimum sensitivity and light collection capability as discussed in (*7*). The system developed by the University of Texas group, POSICAM 6.5, has a smaller deadtime than the larger crystal system of the University of Pennsylvania because the number of independent channels for light detection is large and more collection and processing can be done in parallel. But the problem extends to all of the contemporary systems. What is the optimum detection module or block size? In Monte Carlo simulations, we learn that the block size for BGO is possibly smaller than that used in some systems if deadtime is a major consideration (*7*). The trade-off in light collection and deadtime should not ignore the economic trade-off.

FUTURE DEVELOPMENTS

There are now four or more commercial PET tomographs with spatial resolutions better than 6 mm. Two of these systems, discussed in this issue, have the goal of three-dimensional volume imaging with good resolution and good sensitivity. In both systems, demonstrated performance for axial resolution throughout the volume of <10 mm is still in the development stage, yet the sensitivities and count-rate capabilities and vol-

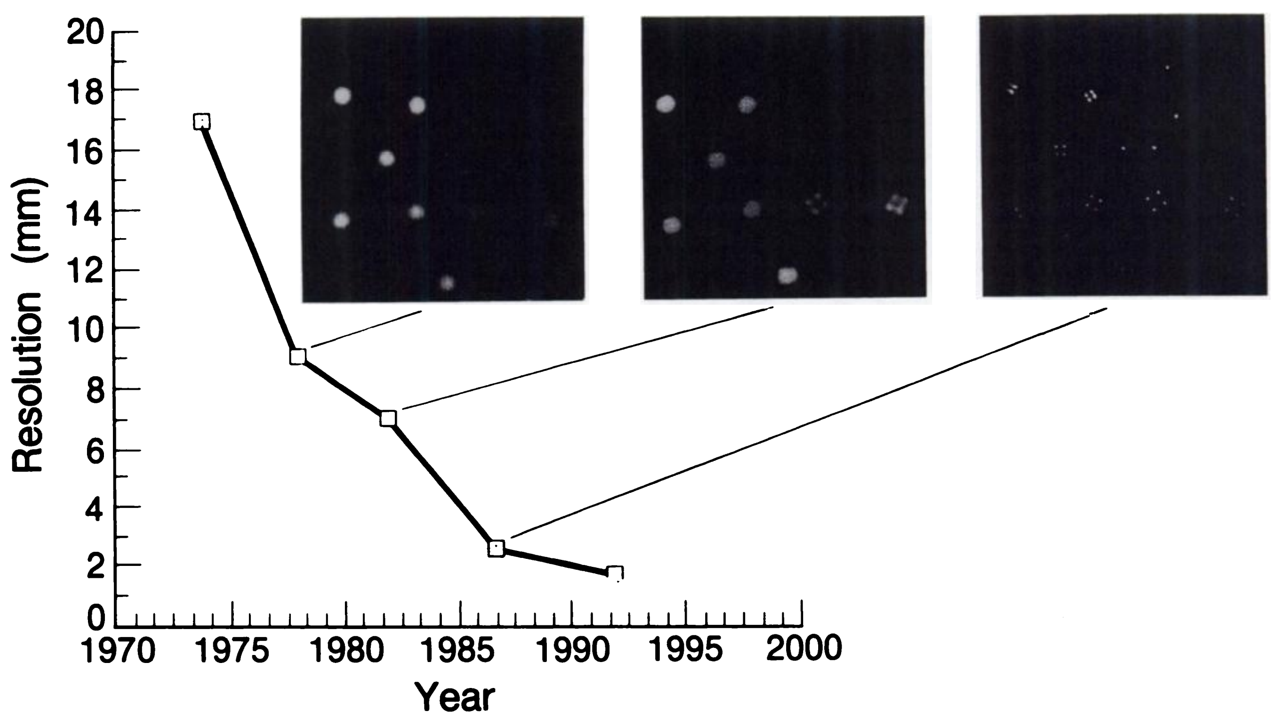

FIGURE 2
The evolution of resolution for PET instruments is shown with the highest resolution of 2.6 mm in the right panel (9). Pattern at nine o'clock is 5 mm.

ume of view make these systems attractive for practical clinical studies. As shown in (*3*), it is possible to improve the sensitivity of SPECT by adding a fourth side to the contemporary one-, two-, and three-sided gamma camera systems. However, the sensitivity of a SPECT system is still much less than that expected or in fact measured for PET systems with similar resolution.

What major innovations can we expect in the future and should we wait before acquisition of a contemporary system? The future expectation for resolution is shown in Figure 2, but we should not wait for a less than 3 mm PET instrument because the required detector system for volume imaging is still under development. Clinical studies in three dimensions are now feasible at improved resolution and this is demonstrated in the articles in this issue of the *Journal*. However, if our objective is research into the balance of neuroreceptors in thin cortical regions of the brain, then we should wait until the required resolution of <3 mm is available. For example, suppose we hypothesize that a disease is based on a disturbance between activity in the putamen and globus pallidus—5.5 mm resolution will be adequate. But suppose we use the same system to investigate the balance of 5HT2 or D2 systems between cortex and striatum. In this suggested study, misleading and falsely disappointing results are foreseen because the instrument is not yet suited to the task. The dream of 2 mm resolution in three dimensions must face the reality that the needed sensitivity will come only if we can cope with the scattered events. As we learned from (*2*), the energy thresholding is effective for scatter rejection if the scintillator has a good light output. The low efficiency of NaI relative to BGO due to differences in stopping power were partly overcome by the improvement in sensitivity by removal of septa. Thus, it appears that the quest for a new scintillator with efficiency of the BGO, the light output of NaI, and the speed of the new scintillator $PbCO_3$ will help make the dream of 2-mm resolution come true.

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