the detectors and the front-end electronics, since the cost of the gantry, computer, and related peripherals remains essentially the same for the two systems.

It has also been our experience that the multislice system is far superior in its clinical applications because of its higher sensitivity and simultaneous acquisition of multiple slices over the organ of interest. It yields several slices in physiologic synchrony, thus permitting a better and easier comparison of information from slice to slice. These factors also facilitate the reordering of the reconstructed images into sagittal and coronal planes. Fast multislice tomographic systems will permit dynamic studies in the clinical situation, whereas a single-slice system would require multiple administration of tracers if more than one slice of the organ is to be studied.

We agree with the authors that a tomographic system should have maximum circumferential efficiency for detection of positrons. However, to achieve this goal we have found that a circular array of detectors exhibits a higher detection efficiency than a hexagonal array for comparable detectors. The hexagonal array offered an easy approach to positron emission tomography, but there is abundant evidence (*2-6*) to indicate that a circular arrangement represents the state of the art in positron tomography. The circular system also offers the physical capability of collecting the data in a shorter period of time than a translate-rotate system. This is an important consideration in a system that may be used for fast dynamic studies, and in minimizing artifacts from patient motion. Redundant sampling in a ring geometry is easily attained and user-controlled by overscanning in the rotation direction.

It is our opinion that successful clinical application of positron emission tomography will be carried out with multislice devices and most likely with circular geometry.

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Reply

The statements in our article (*1*) were not a condemnation of the concept of multislice systems per se. They explain our decision to concentrate on delivery of the best possible images per slice. In order that no one is "mislead," we present here a more complete discussion of the topic by considering each of the points in the preceding letter.

"Hypothetical" five-slice system. This example was given to point out a factor that is particular to computed tomography (CT). In CT, each image is formed by solving an interdependent set of linear equations, and this imposes certain constraints that are not common in medical imaging systems. Any inconsistencies in these interdependent data sets results in a removal of information or addition of artifacts by the reconstruction algorithm. Thus, in the evaluation of different design alternatives for CT, in which there are interdependent data (i.e., data within a plane for each image) and independent data sets (i.e., data for images from one plane to the next), one should try to complete the collection of the interdependent data as quickly as possible. This lessens the probability of errors caused by movement of activity, organ, or patient, or other time-dependent sources of error.

We tried to provide an example in which this point is important, with a five-slice system that had 1/5 the efficiency, per slice, of a single-slice system. With these "hypothetical" systems one could collect five images in the same time. However, if motion or some other disturbance should occur at some time during the study, it would affect only the plane being examined at that time with the single-plane system, but would affect all the planes of the five-plane system. This is a fundamental issue in CT design but it applies, of course, only if the single-plane system is more efficient than *each* plane of a multiplane system. The obvious example cited in the above letter—that of multiple ECAT detector planes compared with a single-detector-plane ECAT—does not apply since it doesn't meet the criteria of the example.

There are many factors involved in CT system design, with different strategies for choosing one set of design criteria over another. This has resulted in a number of different types of positron-emission CT (PCT) systems of which a number are multislice (*2-4*). These multislice systems, when used in a transaxial CT format, were all considered to be similar to the example of the hypothetical five-plane system above when compared with the ECAT i.e., they all have considerably less efficiency per plane than the ECAT. For example, since we feel the PETT IV (*2*) has the highest efficiency per plane among present multiplane devices, it can be chosen as a basis for comparison. We measured the efficiency and resolution of a single pair of PETT IV detectors (ignoring losses caused by position logic of PETT IV) and estimated the efficiency of one plane. Our estimate showed the ECAT to be 4.7 times as efficient as one PETT IV plane. Subsequently Ter-Pogossian et al. (*2*) published the PETT IV efficiency, and the ECAT is now seen to have 5.3 times the efficiency per plane as the PETT IV at comparable resolutions. Thus we do not feel our example was as hypothetical as it was made out to be in the above letter.

Cost. It is very difficult to do true cost accounting in a university setting. We agree that a 30% increase in cost for materials—between, say, the PETT III and PETT IV—may be realistic. However, cost of the added complexity in terms of labor, overhead, service requirements, and maintenance is less well defined in this setting. Yet they are all part of the final product in commercial equipment. Thus we are not in a position to be very confident in final cost estimates.

Clinical applications. In regard to the authors' statement that in clinical applications the multi-slice system is far superior to the single-slice systems, it is our turn to state that this is presented in a fashion both "misleading and incomplete." There are several aspects about the geometry of the PETT IV that the authors have failed to mention. The image planes in the PETT IV are not contiguous. There is about a 3.8-cm gap between detector planes, and therefore in the four-plane configuration the patient had to be moved twice for planes to be contiguous, or moved once in the seven-plane version. This appears to us to limit their point of "physiologic synchrony." The *total* system

TABLE 1.

Configuration	No. of detectors (material)	Detector size	Field-of-view	Resolution*	Efficiency†	Reference	Comments
Circular	64 (NaI)	2×3.8 cm	30 cm	~1.8 cm	4,600	Cho et al. (9)	No. of random and scatter coincidences not given
Circular	95 (NaI)	0.8×2×5 cm	30 cm	~0.7 cm	5,300	Eriksson et al. (10)	No. of random and scatter coincidences not given
Circular‡	280 (NaI)	0.8×3×5 cm	31 cm***	~0.85 cm at center of image ~0.8×1.3 cm at 10 cm from center of image ~0.7×1.9 cm at 15 cm from center of image	11,300	Derenzo and Budinger (11)	Random coincidences have been subtracted and scatter is about 11%
Hexagon	66 (NaI)	3.8×7.6 cm	50 cm	1.7±0.1** 1.3±0.1 0.95±0.1	30,100 15,900 9,200	Phelps et al. (1)	Random coincidence are <1%; scatter radiation is 5.2%

* Resolutions in the axial direction have not been reported to our knowledge for the circular systems. Axial resolution for the ECAT is 1.8 cm with the 1.7-cm inplane resolution, and 1.9 cm with the 1.3- and 0.95-cm resolutions. Significant increases in efficiency should be possible with the Derenzo-Budinger system by making the axial resolution larger.

† Efficiency is for a uniform phantom, 20 cm in diameter, filled with positron activity.

‡ Resolution is estimated with computer simulation of data from a partial set of 16 detectors.

** Resolution throughout the field of view ±1 s.d.

*** This field size is reported in Ref. 11 but field can be larger.

efficiency in the PETT IV (including all seven planes) over the single plane of the ECAT is only modestly higher, and in studies of the brain and heart not all the efficiency is used for the target organ (see later discussion). In view of the gaps between the planes of a PETT IV system, it is not obvious how this device will facilitate the reconstruction of sagittal and coronal planes. In fact, because one probably needs high sampling resolution in the axial dimension to reconstruct properly in the saggital and coronal planes, it is not apparent that the multiplane PETT IV would offer any advantages over the ECAT in this respect. Resolution nonuniformity is also much greater in the PETT IV (2) than in the single-plane ECAT (1).

Fast dynamic studies with tomography are statistically challenging since ECT is itself very demanding from a statistical point of view (5–8). It appears to us that one must maximize the efficiency per plane (i.e., per image) if fast dynamic studies are to become a reality. However, as pointed out in the preceding letter, if this is done in a single-plane configuration, the next plane would require a second administration of tracer. This would also appear to be a problem, although to a lesser degree, in the PETT IV geometry because of the gaps between planes. The efficiency per plane would be limiting, however, from a statistical point of view.

We have chosen to approach the problem of tomographically measuring physiologic functions by sequestering (i.e., trapping), steady-state, or equilibrium approaches that allow tomography of high statistical quality to be performed. However, both dynamic and stationary approaches to the tomographic measure of physiologic function need to be investigated in more detail.

Circular compared with hexagonal tomographs. The above letter states that circular arrays of detectors exhibit higher detection efficiency than hexagonal arrays for detectors of comparable size. However, because of sampling limitations in the circular geometry, small detectors must be used, and these have lower intrinsic efficiency than the larger detectors of the hexagonal design, which has no sampling limitations. In the circular geometry movement (other than 1/2 detector spacing) of the entire detection system, to remove this sampling limitation has not to our knowledge been shown to improve *real* sampling resolution significantly. Our analysis of this motion suggests that sampling can be improved but that there is a sampling nonuniformity that will produce some increase in image noise and a spatially varying resolution in the final image.

The reported values for the efficiency of existing NaI circular ring systems do not support the above claim. These values, shown in Table 1 with the values for the ECAT (1), range from about equal (and this occurs only in a system with four times as many detectors as in the ECAT) to as to low as 1/6 the efficiency of the ECAT at comparable image resolutions.

Circular tomographs do have an attractive feature in that they can collect data without motion of the detectors. This is done, however, at the sacrifice of resolution because of sampling limitations. A small rotation (1/2 the detector spacing) does improve sampling with a modest mechanical complexity. Yet small detectors, with their low intrinsic detection efficiency for 511 keV, must be used to achieve sampling adequate to provide high resolution.

A possible solution to this problem has been suggested by Cho et al. (12) and Derenzo (13): the use of bismuth germanate (BGO) detectors, which outperform NaI in intrinsic efficiency for 511-

keV radiation. Their improved efficiency is particularly apparent in the case of the small detectors required in the circular geometry. BGO detectors are now commercially available and are used in several x-ray CT systems (AS & E, Ohio Nuclear, and Picker). Both the investigators cited are now planning the construction of such devices. Thompson and coworkers (*14*) at Montreal have abandoned their NaI circular-ring tomograph and have constructed and are using a BGO circular-ring system.

An advantage of the circular tomograph in a *stationary* mode is that when activity is moving around during the imaging time, the final image represents a linear superposition (i.e., smooth blurring) of the various redistributions that took place without the "sharp" motion artifacts that are characteristic of CT devices. However, the redundant sampling of the ECAT (*1*) also accomplishes this to a major degree. The redundant sampling of the ECAT has also been shown to provide unique protection against detector instability and motion artifacts, and improves signal-to-noise ratios in the image (*1*).

If a circular tomograph uses detectors that are either stationary or shift half the distance between detectors, a very short imaging time is mechanically possible. It has yet to be proven that the 10-sec scan mode of the ECAT poses any limit in mechanical scan times that exceeds the limits imposed by statistical requirements.

We would like to reiterate that our comments (*1*) were not against a general design concept of a multiplanar ECT system. A *properly* designed ECT system will ultimately be limited in resolution and contrast by its detection efficiency. As opposed to x-ray CT, which irradiates mainly the plane under study, all the potential planes in ECT are being irradiated whether they are imaged or not. It seems only logical, therefore, to try to use this information. The problem is not easily solved, however, and different design options must be carefully analyzed, and tradeoffs made, to optimize the overall system performance within realistic design constraints, considering the problems particular to CT and the types of studies for which the system will be used.

When one examines the parameter of efficiency he must consider: a) the total system efficiency; b) the efficiency per plane (or image); and c) efficiency for the organ under study. The system efficiency tells how efficient the system is when activity covers the entire field of view. The per-plane efficiency gives the efficiency per image. The last factor tells how much of the system efficiency will be used to image the organ of interest. Present multiplane imaging systems employ only 40–70% of the *system efficiency* for heart and brain studies, since the remainder of the planes are outside the organ of interest. A single-plane system, on the other hand, inherently has a design objective of maximizing the efficiency both per image and for the target organ. In addition, multiple-plane systems must be carefully designed to avoid gaps between planes [the multiplane system of Muehllenhner et al. (*4*) has continuous or redundant planes], or to provide appropriate spacing (by moving the patient) to allow interplane images to be recorded in a contiguous fashion, or with some appropriate overlap, to optimize axial sampling (i.e., to minimize partial-volume effects). Along with efficiency values, one should provide that portion caused by scatter and random coincidences, since these events provide counts but little or no information. The resolution at the stated efficiency should also be given.

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Optimization of Analog-Circuit Motion Correction for Liver Scintigraphy

Baimel and Bronskill (*1*) have presented an interesting mathematical proposal, though whether it will significantly improve the accuracy of reading liver scintigrams in clinical practice is another matter. Their sweeping conclusion that *all* cameras should be provided with analog motion correction is a little premature, particularly when their paper does not contain even a solitary pair of liver scintigrams—with and without their motion correction. This is in spite of their studying 52 patients and the claim that in seven cases (13%) the clinical interpretation was changed. Moreover, they do not state whether any of the 52 scintigram reports were confirmed by either operative or autopsy findings. Also, it appears from their text (p.1064) that the clinical evaluation was made by a single nuclear medicine