THE AUTHOR'S REPLY

Levy quite correctly points out that the gain in signal-to-noise ratio (SNR) to be expected when using a Fresnel zone plate is dependent on the nature of the source intensity distribution. Only for a single point source can the large collection efficiency of the zone plate be translated directly into a corresponding reduction in dose or exposure time. This fact was discussed in our earliest publications (*1–3*) on this subject, and a simplified derivation of the SNR has recently been published (*4*). A very detailed treatment of quantum noise in zone plate imaging, taking full account of the spatial distribution of the noise field and the limited spatial bandwidth of the reconstruction system, has been submitted for publication (*5*).

Although Levy's mathematics is much oversimplified (most of his equations are not even dimensionally correct), his approach is sound. The result that the SNR gain g is related to the ratio of the intensity at the point of interest to the average inensity of the source is also correct. An equivalent statement is given in Ref. *3*. However, it definitely does not follow from this that "the nature of the image and the origin of the noise in nuclear medicine does not generally enable us to use the advantages" of the zone plate or that the "gain in SNR will be significant only for the point of the image far superior to the average of the image."

The difficulty here is in specifying just how this "average of the image" is to be computed. In fact, although Levy does not put limits on his integrals, they should be performed over a region approximately equal to the zone plate shadow (*5*). For example, with a 5-in.-diam zone plate used with a 10-in. diam Anger camera and unit magnification ($s_1 = s_2$ in the notation of Ref. *2*), the average must be taken over a 10-in. disk. For a uniform flood source of this size or larger, each point will have the same intensity as the average. The gain g will then be a little less than one and there will be a slight disadvantage to the zone plate compared with the equivalent pinhole. At the other extreme, for a point source, g is approximately equal to the collection efficiency advantage and can be as large as a thousand or so (provided, of course, that only quantum noise is present).

Real clinical situations usually lie between these two extremes. Perhaps liver and lung imaging, where the object nearly fills the field, approach the flood source limit, but certainly in bone, thyroid, and kidney imaging the area of the object is small compared with the area of the zone plate shadow and there will be a significant advantage to the use of a zone plate.

On the other hand, the overall usefulness of the zone plate should not be assessed on the exposure time advantage alone. In our laboratory we have been using the zone plate primarily with x-ray film as the detector (*3,6*). This combination is substantially *slower* than an Anger camera and collimator but offers advantages in resolution, simplicity, and portability and, as noted by Levy, tomographic capability. On the negative side, the photographic and optical processing is still somewhat tedious and time-consuming and there is the possibility of artifacts in the image (*3*). This camera has been used successfully in a variety of clinical studies (*7*) and is indeed capable of imaging large organs (*8*).

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THE AUTHORS' REPLY

We greatly appreciate Dr. Levy's comments regarding the signal-to-noise (SNR) in coded aperture imaging and agree with his comments regarding the gain in SNR for strong sources with corresponding loss for weak sources.

In order to treat the problem completely, one must explicitly include the aperture code and source distribution as Dr. Levy indicates. Figure 1 gives some preliminary results of such an analysis performed for a stochastic aperture (*1*). The error kernel, E, normalized to the peak value of the point response function is shown plotted as a function of distance

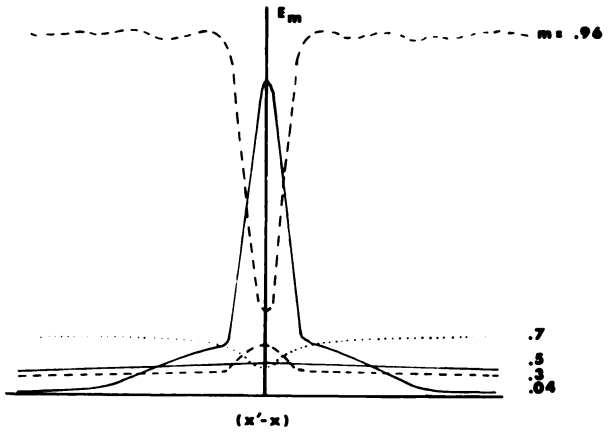

FIG. 1. Plot of error kernel as a function of displacement for stochastic aperture of various mean transmission, m.

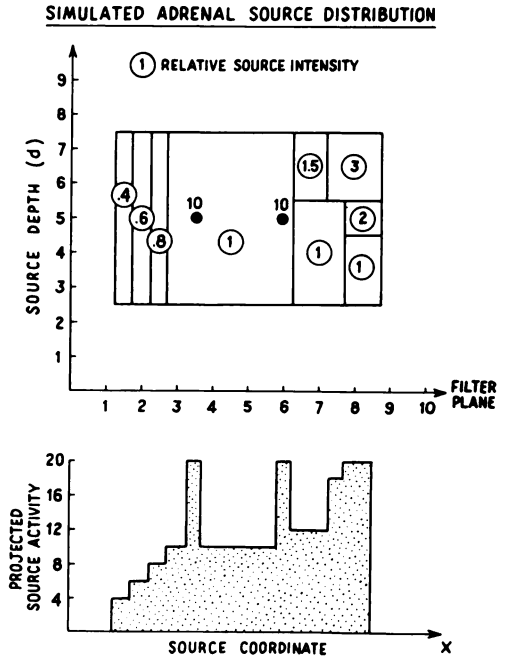

FIG. 2. Source distribution utilized for digital simulation of adrenals. The detector is 10 units wide and located 4 units below filter or aperture plane.

for stochastic apertures of various mean transmission, m. In order to obtain the error distribution in the image, the object distribution must be convolved with the error kernel. One can see that as m gets small, approaching the pinhole as a limit, the variance increases but is spatially localized. As the mean transmission is increased, the error is locally reduced but increased over the rest of the field. For m = 0.5 the distribution is uniform and closely approximates the case of a zone plate which fills the detector field.

There are many instances in nuclear medicine in which the image does not fill the detector field efficiently. Since the average image intensity should be determined as an average over the entire detector,

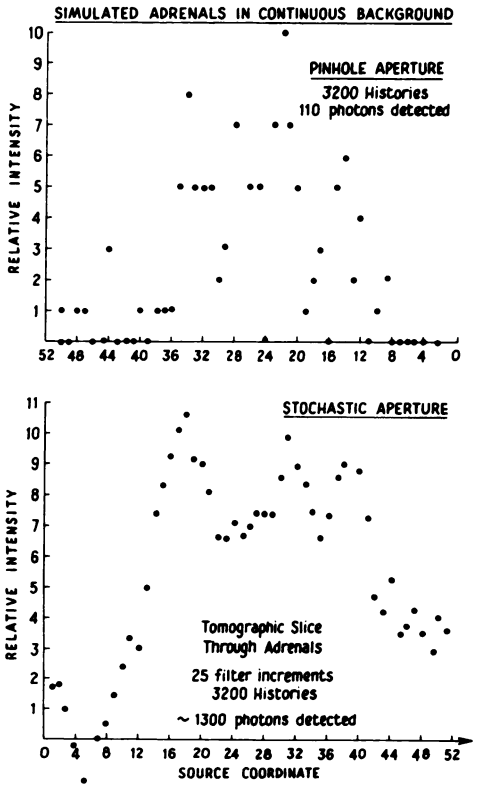

FIG. 3. Reconstructed images of adrenal using pinhole aperture (top) and stochastic aperture (bottom).

there will be studies in which an appreciable gain in signal-to-noise will be realized. Among these are blood flow, skeletal, and small organ imaging. By way of illustration, a two-dimensional digital simulation has been performed for the case of a pair of adrenal glands. A continuous source distribution having width and depth was "viewed" with equal resolution and for equal time with a pinhole and a stochastic aperture of m = 0.5. The source distribution is shown in Fig. 2. Two adrenals are depicted in a background which increases to the right to simulate the liver. The pinhole image is shown at the top of Fig. 3 where the relative intensity is equal to the detected counts in each detector element. The image reconstructed from the stochastic aperture is shown at the bottom. The calculated error is approximately uniform across this image (from Fig. 1) and is equal to 1.5 relative intensity units. The error in the left adrenal peak height is 13.9%. The pinhole data which must be averaged over three detectors shows an uncertainty of 23.5%. In order to achieve comparable accuracy the pinhole exposure would need to be increased a factor of 2.9. Further improvement could be obtained by optimizing the mean transmission of the aperture for this source distribution.

The above example is somewhat limited and serves to illustrate only one aspect of coded aperture imag-

ing. The overall significance of coded aperture imaging is the added freedom which the technique can bring to the solution of nuclear medicine imaging problems. For instance, the relationship between aperture transmission and the shape of the error kernel illustrated in Fig. 1 provides a flexibility which allows some control over the error distribution in the image. The problem of detector uniformity becomes greatly reduced with coded apertures since the coded image is spread over a large area of the detector. The advantage of coded apertures cited by Dr. Levy and others (*2*) in the case of detector noise permits, for example, consideration of gamma-ray detectors utilizing large diameter, efficient photocathodes, with correspondingly increased thermal noise. Or, film may be used as a detector in spite of its intrinsic fog level and the small number of film grains exposed by an interacting gamma-ray (*3*). One also has the option of designing apertures with spatial frequency response functions tailored to specific needs.

These are some of the reasons we feel that coded aperture imaging stands an excellent chance of bringing significant advances in nuclear medicine and will provide a fruitful area of research for some time to come in spite of the fact that the gain in imaging efficiency does not simply equal the gain in solid angle.

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ACKNOWLEDGMENT

We wish to express our gratitude to A. Z. Akcasu, R. S. May, and G. F. Knoll of the University of Michigan, Department of Nuclear Engineering, for the use of their preliminary results for the stochastic aperture.

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A CODING SYSTEM FOR NUCLEAR MEDICINE USING UNIT RECORD EQUIPMENT

In 1969 we designed a coding system to fit our special needs. We feel it has worked so well for us that it may be of interest to others. It uses code cards, computer punch cards, a two-digit filing system, an alphabetic card file, and unit record equipment, i.e., an IBM 407 accounting lister, an 082 card sorter and an 026 card punch, for processing the files. The 5 × 8 in. code card serves as an alphabetic cross-file card. It is imprinted with groups of rectangular blocks representing the 80 columns on a punch card as shown in Fig. 1, purposely arranged in four rows.

The receptionist fills in the top row with the patient's personal information: social security number, last name, initials, and last two digits in the year of birth. If the name is longer than 10 letters, the tenth letter is replaced by an asterisk. The card with this information is attached to the study request and given to the technologist.

The second row is completed by the technologist and contains study or test information. Space is provided for entering the date of the test, the amount of radioactivity and its unit (mCi or μCi), the route of administration, the procedure, and the instrument. The isotope is identified by its mass number. Mnemonic code characters are used in this row so they can be remembered easily but a listing of the code is kept readily available. Below are examples of the study code.

SB—scan, brain
SBF—scan, brain and cerebral blood flow
STU—scan, thyroid, uptake
RG—renogram

When the procedure is completed, the request with the attached card and the actual technical data such as scans, recording, and calculations are given to the physician.

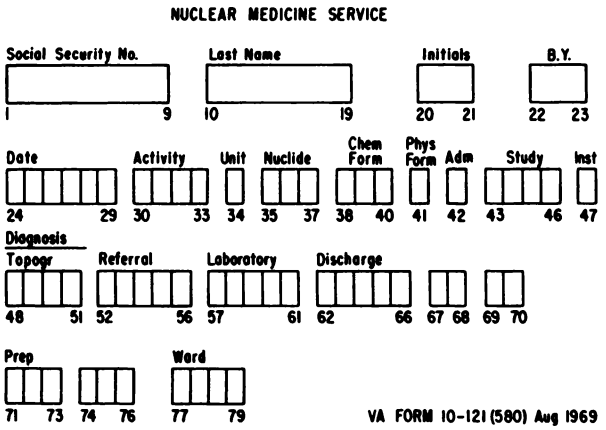

FIG. 1. Code Card. Note grouping of information in four rows: First row, patient's personal data (clerk), second row, description of work done (technician), third row, diagnostic information (physician), fourth row, additional data.