statistical image degradation was encountered at clinical counting rates. This phenomenan can easily be explained when it is realized that the time required to get equal statistics is not governed alone by the measured counting rates as Bloch and Sanders imply. This can be shown by the following argument. Let C_1 be the counting rate in the 125–170-keV window, C_2 be the counting rate in the 91–102-keV window, and C_3 ($= C_1 - C_2$) be the final counting rate with the scatter-subtraction technique. Then the percent standard deviations in the counting rates for the unsubtracted case ($\sigma C_1/C_1$) and the subtracted case ($\sigma C_3/C_3$) are

$$\sigma C_1 C_1 = 1\sqrt{C_1 t_1} \quad (1)$$

$$\sigma C_3/C_3 = \sqrt{C_1/t_2 + C_2/t_2/}(C_1 - C_2), \quad (2)$$

where t_1 and t_2 are the respective counting times. Equating Eq. 1 and 2 to give equal signal-to-noise ratios, we find that

$$t_2 = [C_1(C_1 + C_2)/(C_1 - C_2)^2]t_1. \quad (3)$$

Hence, using Bloch and Sanders' data, we calculate that the counting time using scatter subtraction must be 197% greater than that for the 125–170-keV window. Bloch and Sanders state that the counting time must be increased by only 50%. Using these same arguments, one can see that even the 140–170-keV scan will give less statistical fluctuations than the scatter-subtraction method. Since this window has about the same MTF, it will produce a better scan.

We have also found experimentally that a system with a high baseline will always give the same or better scan than the scatter subtraction technique if the statistical level is the same. The reason why the scatter subtraction method appears to work in the phantom Bloch and Sanders have scanned is that the count density is so high (25,000 counts/in.^{2}) that statistical fluctuations are no problem.

To strengthen the argument, Beck has shown that in order to obtain the best image, some small-angle scatter is desirable (*3*). A variable baseline system uses this principle to reduce the effects of scattered radiation (*4*). The best baseline is found for each scatter to primary ratio. Therefore, not only is scatter rejected, but an optimum statistical level is maintained.

LARRY W. OBERLEY
STEVEN C. LENSINK
JAMES C. EHRHARDT
University of Iowa
Iowa City, Iowa

REFERENCES

1. BLOCH P, SANDERS T: Reduction of the effects of scattered radiation on a sodium iodide imaging system. *J Nucl Med* 14: 67–72, 1973

2. EHRHARDT JC, OBERLEY LW: Effects of spectral changes on scanning. *Radiology* 104: 207–208, 1972

3. BECK RN, SCHUH MW, COHEN TD, et al: Effects of scattered radiation on scintillation detector response. In *Medical Radioisotope Scintigraphy,* vol 1, Vienna, IAEA, 1969, pp 595–615

4. OBERLEY LW, EHRHARDT JC, LENSINK SC: The variable baseline scanner. *Phys Med Biol* 17: 630–637, 1972

AUTHORS' REPLY

Oberley, Lensink, and Ehrhardt believe that the perceptibility of a noisy object in a noisy background is the same when the signal-to-noise ratio, SNR, is the same in the two images. This is an interesting approach but we believe that they did not carry it to its conclusion.

A more complete calculation of the SNR is as follows:

Let C_t and C_0 be the counting rate in the 125–170-keV window over two regions of a radioisotope distribution to be distinguished, and C_s the counting rate in the scatter window, 91–102 keV. To simplify the algebra, assume that the counting rate in the scatter window is the same over the two regions. This will very nearly be true for threshold images with low contrast. Without scatter subtraction, the signal is $(C_t - C_0)$ and the noise due to random fluctuations in the signal is

$$[(C_t + C_0)/t_1]^{1/2}$$

where t_1 is the measuring time.

The signal-to-noise ratio is

$$SNR = (C_t - C_0)/[(C_t + C_0)/t_1]^{1/2}. \quad (1)$$

When the counts in the scatter window are subtracted from the counts in the photopeak window the signal-to-noise ratio can be written

$$SNR = \frac{(C_t - C_s) - (C_0 - C_s)}{[\{(C_t + C_s) + (C_0 + C_s)\}/t_2]^{1/2}} \quad (2)$$

where t_2 is the counting time.

The counting time for the same signal-to-noise ratio using the scatter-subtraction technique is obtained by equating Eq. 1 and 2,

$$t_2 = t_1[1 + 2C_s/(C_t + C_0)]. \quad (3)$$

For 10 cm of scattering material our measurements indicated that $C_s = 0.33C_0$, and for C_t 20% greater than C_0, for example, the counting time using the scatter-subtraction technique should be increased 30%.

There are, it seems to us, serious shortcomings in using the signal-to-noise theory as formulated above as a criterion for image detection. It errs in treating counts due to photons that have been scattered as random noise. These counts are deleterious because they degrade image contrast. By subtracting them, the image quality is improved even though, with equal scanning time, the counts per picture element decreases somewhat so that the image appears to be noisier. We believe that Fig. 7 in our paper illustrates this point.

Another approach to image detection was suggested by Rose (*1*). He demonstrated, for a wide variety of perception data in visual physiology, that image perceptibility is the same when the standard deviation in the number of counts accumulated in the object is the same in the two images. Sturm and Morgan (*2*) applied this model to problems of perception in radiology. They showed, for example, that the relationship between the diameter of the smallest discernible object, δ, and its contrast, C, is

$$\delta = \frac{2k}{C}\left(\frac{g}{\pi t n_0}\right)^{1/2} \times 10^2$$

where g is an amplification factor which would be unity in radionuclide scanning, n_0 is the number of photons/mm^2, and t is the accumulation time during which the image is formed. The constant k is determined experimentally. Thus two images will be equally perceptible, considering only quantum statistics if the products $(n_0 t)^{1/2}$ are equal. This is the approach we used in discussing the data of Table 1 in our paper. In our example, the subtraction technique reduced the system sensitivity to n_0 by a factor of 0.67 (with 10 cm of scattering material) compared with a conventional spectrometer setting of 125–170 keV. Thus it is necessary simply to increase the counting time t by a factor of $1/0.67 = 1.50$ to regain the same image perceptibility.

Thus, according to this approach, the scatter-subtraction method requires only a moderate increase in scan time in exchange for an increase in image contrast.

Finally, the count density in the photoscan of a phantom of 4,000 counts/cm^2 or 25,000 counts/$in.^2$ shown in Fig. 7 of our paper is not considered clinically unreasonable. In our department we routinely control the accumulation time on our Picker Dyna Camera for recording the radioisotope distribution in a brain study by adjusting the count density to be 1,000 counts/cm^2 over a region in the brain containing a small amount of activity. There are certainly regions in the image that contain more than 4,000 counts/cm^2.

PETER BLOCH
TOBY SANDERS
Hospital of University of Pennsylvania
Philadelphia, Pennsylvania

REFERENCES

1. ROSE A: Sensitivity of the eye on absolute scale. *J Opt Soc Am* 38, 196, 1948

2. STURM RE, MORGAN RH: Screen intensification systems and their limitations. *Am J Roentgenol* 62, 617, 1949

COMPARATIVE STUDY OF INSTANT TECHNETIUM FROM VARIOUS COMMERCIAL SUPPLIERS

The use of instant ^{99m}Tc produced by methyl-ethyl-ketone extraction or by some alternate means offers several advantages over $Na^{99m}TcO_4$ obtained from chromatographic generators. Instant technetium is less expensive and saves time that the technologist could allocate to other tasks. It also reduces the radiation dose to the technologist because of the absence of exposure to a ^{99}Mo-containing generator. In addition, instant technetium has been shown to have decreased amounts of radioisotopic and radiochemical impurities. In particular, ^{99}Mo was said to be at a level of 1/100 of that found in chromatographic generators (*1*).

This communication presents the results of a comparative study of instant technetium made available by four commercial suppliers with regard to ^{99}Mo concentrations, utility of ^{99m}Tc concentration, and adequacy of container shielding.

Molybdenum-99 was determined by a technique consisting of placing a syringe containing a 1-ml sample of ^{99m}Tc eluate into a ¼-in. lead-shielded stand which fits over a well scintillation counter. The activity of the sample was recorded by integral counting and compared with the activity of a calibrated ^{99}Mo standard counted in the same geometry. A small correction factor for the incomplete attenuation of ^{99m}Tc radiation through the ¼-in. lead was necessary to be subtracted from the measured ^{99}Mo counts of the eluate. This correction factor was easily determined by counting an initial ^{99m}Tc eluate sample at time of arrival in our laboratory and a few days later when there was no ^{99m}Tc radiation remaining. Table 1 shows the results of one week trials of ^{99m}Tc from five different sources.

The finding of significant levels of ^{99}Mo activity in the eluates of commercially supplied instant tech-