tration of the activity can be described by the following equation:

$$A(t) = 970\, e^{-0.0294t} + 30\, e^{-0.0009t} - 1000\, e^{-0.0875t}\, \mu Ci$$

The total cumulated activity ($\tilde{A}$) in the brain is calculated to be 54,880 μCi-hr. The volume of CSF in the head region is assumed to be 130 ml, the absorbed fraction, ϕ, to the surface of the surrounding tissues from particulate radiations is assumed to be 0.5 and $\Sigma_i \Delta_i \phi_i$ is 0.1426. For the penetrating radiation it is assumed that the radioactivity is uniformly distributed in the brain tissues and $\Sigma_i \Delta_i \phi_i = 0.127$.

Based upon these assumptions and parameters, the total radiation dose to the surface of the brain tissues in contact with CSF in the ventricles and posterior fossae is calculated to be 65 rad/mCi administered activity. The average dose to the brain from penetrating radiation only is calculated to be 5 rad/mCi. The uncertainty in biologic data based upon small numbers of patients is large but the values agree with the data presented by Morris and DeLand and the radiation dose is an order of magnitude less than reported by Barbizet, et al (*1*).

Although it does not appear that prolonged retention in the meninges takes place under usual circumstances, it is possible the observations of Barbizet, et al may be related to some pathologic or anatomic variant yet unrecognized. We therefore suggest further evaluation and corroboration of the safety of this agent for cisternography.

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REFERENCES

1. BARBIZET J, DUIZABO P, THOMAS J, et al: Danger of ytterbium-DTPA in isotopic cisternography and ventriculography. *Nouv Presse Med* 1: 2899–2901, 1972

2. MORIN RL, DELAND FH: Safety of ^{169}Yb-DTPA in cisternography. *J Nucl Med* 15: 375–376, 1974

3. OLDENDORF WH, IISAKA Y: Interference of scalp and skull with external measurements of brain isotope content: Part 2. Absorption by skull of gamma radiation originating in brain. *J Nucl Med* 10: 184–187, 1969

EFFECTS OF SCATTER SUBTRACTION ON IMAGE CONTRAST

It appears that some confusion exists regarding the improvement in image contrast described by Bloch, et al (*1*) as evidenced by Inia's Letter to the Editor (*2*) and by the author's response.

Bloch and Sanders used a subtraction technique intended to compensate for the contrast reduction due to Compton-scattered photons which produce pulses that occur within the photopeak window of a NaI(Tl) detector system. Even with an "optimum" baseline setting of 126 keV for a large uniform volume distribution of ^{99m}Tc, the window set on the photopeak will contain a significant scatter fraction (*3*). The scattered photons give rise to the characteristically long tails of the line-source response functions (LSRF) measured in a scattering medium. The effect of these tails is a reduction in image contrast.

If a second window could be set somewhere on the scatter spectrum so as to produce a LSRF having the same magnitude and shape as the scatter component within the photopeak, then subtraction of the former LSRF from the latter would eliminate contrast degradation due to scatter pulses in the photopeak window. The principal concern in the original article, however, was only with the magnitude of the scatter response; that is, the shape of the scatter response was not treated explicitly. To compensate for this magnitude, Bloch and Sanders set a second window with the baseline at 91 keV near the backscatter energy. The width of this window was adjusted to yield approximately the same number of scattered photons as were contained within the photopeak window. This setting was based on the assumption that the Klein-Nishina equation adequately describes the observed pulse-amplitude spectrum due to scattered photons. The events occurring in the scatter window were then subtracted from those in the photopeak at each position in the scan. Although this approach may compensate accurately for the magnitude of the scatter component in the photopeak, image contrast is improved only because the shapes of the line-source response functions due to scatter were approximately the same for the two windows; therefore, this procedure produced a net line-source response function with reduced tails compared with the photopeak window alone. The reduction in the tails results in an increase in the system MTF(ν)* at all spatial frequencies, ν (cycles per unit

* The MTF(ν) is the magnitude of the detector transfer function which is the Fourier transform of the line-spread function (*4*).

distance), and hence leads to the increase in contrast observed by Bloch and Sanders.

These points are illustrated in Figs. 1 and 2, obtained with ^{141}Ce (145 keV) and a Ge(Li) detector, with windows set to simulate the conditions and assumptions of Bloch and Sanders. Figure 1 shows that the response in the scatter window exceeds that in the photopeak window when the line source is at a distance greater than approximately 1 in. from the detector axis; therefore, subtraction yields a bipolar response function with negative tails (dashed curve). This overcompensation for scatter results in rising values of MTF(ν) for $\nu > 0$ as seen in Fig. 2; thus, in the frequency range where MTF > 1, the loss of contrast due to scatter in the photopeak window is more than regained by the subtraction of events in the scatter window. Since MTF(ν) can be interpreted as the efficiency with which the detector is able to transfer modulation or contrast at each spatial frequency from the object to the image, the consistently higher values of MTF resulting from the scatter subtraction imply an increase in image contrast.

In summary, we feel that the following points should be emphasized:

1. For a line source embedded within a scattering medium, the response due to scattered photons in any given energy range will have a definite average amplitude and shape for a given set of imaging parameters.
2. A scatter subtraction technique will yield a net line-source response function. It is proper to compute a detector system MTF from this function since the method preserves the properties of a linear operator and (neglecting effects near the edges) produces a shift-invariant response function.
3. The cancellation of the tails will not, in general, be precise because the shape of the scatter response varies with energy.
4. The subtraction technique should not be confused with "background erase." The magnitude of the scatter response subtracted at each position in a scan is a function of the distribution of the activity within the object whereas with background erase a constant level is subtracted over the entire image. Although this also increases the contrast, it cannot be included in the system MTF because it is a nonlinear operation.
5. Although both scatter subtraction and background erase increase the image contrast, it is important to recognize that both of these operations reduce the signal-to-noise ratio in the resultant image. Therefore, they may be used to some advantage only when the signal-to-noise ratio is well above the threshold of detectability for structures of interest.
6. On the other hand, the signal-to-noise ratio can be maximized if the photons at each energy are allowed to contribute to the image with a weighting factor that is proportional to the contrast which such photons can provide for the structures of interest (*5,6*). These weighting factors are generally positive and result in reduced contrast but increased signal-to-noise ratio compared with the image formed by photopeak events alone.
7. Finally, we note that Eq. 3 of the original article was misprinted and should be:

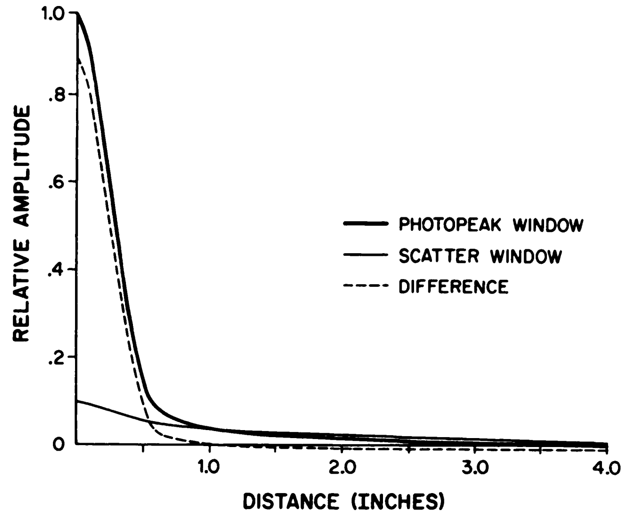

FIG. 1. Normalized line-source response functions for ^{141}Ce (145 keV) measured in focal plane (10 cm) with 8 cm of overlying scattering medium. Analyzer settings were 130–170 keV for photopeak window and 93–110 keV for scatter window.

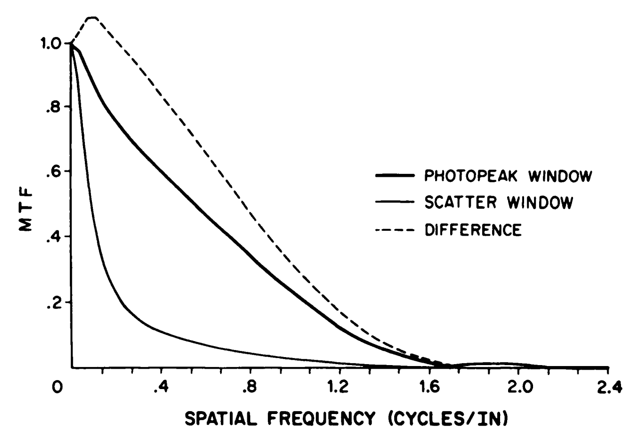

FIG. 2. MTF calculated from response functions of Fig. 1. Curve representing scatter subtraction is uniformly greater in amplitude than that associated with photopeak window alone; hence, image contrast would be increased by this technique.

$$N = \frac{\pi r_0^2 m_0 c^2 n}{E_0^2}\left[\frac{E_1^2 - E_2^2}{2E_0} + (E_1 - E_2)\left(\frac{2m_0c^2}{E_0} + \frac{m_0^2c^4}{E_0^2}\right) + \ln\left(\frac{E_1}{E_2}\right)\left(E_0 - 2m_0c^2 - \frac{2m_0^2c^4}{E_0}\right) - \left(\frac{1}{E_1} - \frac{1}{E_2}\right)m_0^2c^4\right].$$

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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. Inia P: Reduction of the effects of scattered radiation. *J Nucl Med* 15: 316, 1974

3. Beck RN, Schuh MW, Cohen TD, et al: Effects of scattered radiation on scintillation detector response. In *Medical Radioisotope Scintigraphy*, Vienna, IAEA, vol 1, 1969, pp 595–616

4. Beck RN: Nomenclature for Fourier transforms of spread functions of imaging systems used in nuclear medicine. *J Nucl Med* 13: 704–705, 1972

5. Beck RN, Zimmer LT, Charleston DB, et al: Advances in fundamental aspects of imaging systems and techniques. In *Medical Radioisotope Scintigraphy*, Vienna, IAEA, vol 1, 1973, pp 3–45

6. Beck RN, Zimmer LT, Charleston DB, et al: A theory of optimum utilization of all detected radiation. In *Proceedings of the University of California Symposium on Semiconductor Detectors in Medicine, San Francisco, California, March 1973*, Oak Ridge, Tenn, USAEC, CONF-730321, 1973, pp 87–106