gram was evidently not subdivided according to subject matter, the subject index which is well cross-indexed lets the reader scan the volume according to his interest. The list of participants provides a ready source of reference.

Abstracts are presented in English, French and German, and the Discussion following each presentation provides additional sources for active or interested participants.

Physically, the volume is very appealing and well assembled and should provide a ready source for frequent reference.

The editors and publishers are to be commended for making the Proceedings available so quickly.

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nm/ LETTERS TO THE EDITOR

A PRACTICAL FIGURE OF MERIT IN RADIOISOTOPE IMAGING

I read with interest the paper "Relative Importance of Resolution and Sensitivity in Tumor Detection" by Westerman *et al* in the *Journal of Nuclear Medicine* (*J. Nucl. Med.* **9**:638, 1968).

May I suggest that these writers' experimental findings may be conveniently formulized using a rather simple theory which, although rather philosophically conceived, provides a very usable, practical, comparison criterion for radioisotope systems?

In analyzing the results of imaging procedures, we perceive useful information through changes in dot concentration (or film density). Thus a system that provides a higher "image density" (e.g. counts/unit area for a subject point source) will be more efficient at displaying real differences because of enhanced statistics. A scintillation camera with point-source sensitivity, S, will thus "smear" the image dots from each subject "point" over an area proportional to R^2, where R is the f.w.h.h. resolution distance, such that we may define a simple figure of merit, proportional to image density, as

$$FM_1 = S/R^2. \qquad (1)$$

In situations where point-source sensitivity is invalid (e.g. for focusing-collimator scanners), the plane-source sensitivity and resolution distance for a given collimator-to-subject distance may be substituted with equal validity in Eq. 1.

To take exposure time T into account, all we need do is realize that from a count or image density point of view doubling in exposure is equivalent to doubling in sensitivity, etc. Thus our overall figure of merit is given by

$$FM = ST/R^2. \qquad (2)$$

Therefore to compare two scintillation-camera systems (different collimators) with the assumed same *shape* of point-source response functions and for the same application all that is required is to compare FM values. It is interesting to note that this figure of merit contains no reference to subject size, concentration ratios, etc., but provides a *relative* detection probability criterion. So when comparing System 1 with System 2, the merit ratio MR is given by

$$MR = FM_1/FM_2 \qquad (3)$$
$$= (S_1/S_2)(T_1/T_2)(R_2/R_1)^2. \qquad (4)$$

To consider Westerman's first experiment of imaging bulbs with equal exposure time and taking the "technetium collimator" as System 1 and the "coarse collimator" as System 2, we have

$$S_1/S_2 = 1/4$$
$$T_1/T_2 = 1$$
$$R_2/R_1 = 2.2/1.5.$$

Substituting into Eq. 4, and evaluating the merit ratio, we get

$$MR = 0.54.$$

Thus, in spite of its coarser resolution, System 2 has an increased probability of detecting abnormalities because of its greatly superior sensitivity.

Westerman notes that while the coarser-resolution system requires an exposure time of 13 sec, the other system requires 25 sec for approximately equivalent perception. This agrees with the theory presented, since

$$T_2/T_1 = 0.52,$$

which produces an MR value in Eq. 4 of virtually 1.0; i.e. both systems, under these conditions, have equal detection merit.

The fact that a reduction in resolution area by a factor 2.2 $[(R_2/R_1)^2 = 2.2]$ requires an increase in counting time by a similar factor $T_1/T_2 \simeq 2.2$ is particular to these two systems with sensitivity ratio $S_1/S_2 = 4$ since on substituting into Eq. 4 we get

$$\begin{aligned} MR &\simeq \tfrac{1}{4} \times 2.2 \times 2.2 \\ &\simeq 1.0 \end{aligned}$$

which satisfies our equal perception probability criterion. In general, $T_1/T_2 \neq (R_2/R_1)^2$ for equal perception, the required value of T_1/T_2 depending on both $(R_2/R_1)^2$ *and* S_1/S_2.

Turning now to Westerman's clinical example and taking the data from the text (which unfortunately varies slightly from that given in the figure caption), we have:

$$\begin{aligned} S_1/S_2 &= \tfrac{1}{4} \\ T_1/T_2 &= 2/1.5 \\ R_2/R_1 &= 2.2/1.5. \end{aligned}$$

Again substituting into Eq. 4 we get a merit ratio MR of 0.73 which then indicates System 2 (coarse collimator) to be superior in detection ability, as borne out by Westerman's experimental findings.

In summary, Eq. 2 affords us a very simple figure of merit for radioisotope imaging systems which can be almost mentally applied in practice. It is particularly useful when considering the use of a high-resolution collimator to attempt to image detail since a simple comparative test using Eq. 4 tells us how much more exposure time is required, compared with a coarser collimator, just to obtain the same detection probability.

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A REPLY: DETECTING ABILITY OF RADIOISOTOPE IMAGING DEVICES

We are interested in Dr. Walker's suggestion of the factor S/R^2 as a simple figure of merit related to detecting ability (R and S are the resolution diameter f.w.h.h. and sensitivity for point sources). His concept is entirely in agreement with—although less exact in its relationships than—the concept developed in a full analysis by one of us (Sharma, 1969).

The parameter that is proportional to S/R^2 is found to be the "detecting ability, D," or reciprocal of the time required to achieve a given level of statistical confidence (say n standard deviations) in the expression

$$n = \frac{\text{Difference in no. of counts between "suspect" area and "normal" area}}{\text{Standard deviation of this difference}}. \quad (1)$$

The full equation relating detecting ability to the other parameters is, for a Gaussian point spread curve, given by:

$$D = \frac{\mu v^2 C_N S\,(1 - p/100)^2\,(f - 1)^2\,e^{-2\mu d}}{n^2[v(1 - p/100)(f - 1)\,e^{-\mu d} + 1.44\,\pi R^2\,(\log_e 10 - \tfrac{1}{2}\log_e p)(1 - e^{-\mu l})]} \quad (2)$$

in which

- D = reciprocal of time required to achieve a critical level of significance (i.e. any given value of n in Eq. 1)
- v = volume of source to be detected
- f = ratio of concentration in source to that in the surrounding medium
- C_N = concentration of radioisotope in surrounding medium
- μ = total linear attenuation coefficient of the photons by the medium
- d = depth of source in medium
- l = thickness of patient or phantom
- p = percent isocount contour defining the test area.

At the condition of threshold detection, the second term in the denominator (representing background counts from the surrounding tissue) is predominant, and $D \alpha v^2 C_N S/R^2$, for given values of f, μ, d, p, l and n.

Walker's factor S/R^2, being proportional to D in certain conditions and thus inversely proportional to the critical time required for threshold detection of a small source, is indeed useful. It can of course only be used for limited comparisons under comparable conditions unless the coefficient in Eq. 2 is evaluated. A corresponding equation has been derived for a triangular point spread curve (Sharma, 1969).

These formulations are quite valid for both moving (e.g. rectilinear) and stationary (e.g. camera) imaging devices, but not for the stationary focusing-collimator head of a moving scanner; Walker's letter is not clear on this point. Detecting ability is, however, only proportional to S/R^2 for point sources or for sources smaller than the resolution diameter R.

We have also considered the variation of R^2/S