JOURNAL OF NUCLEAR MEDICINE 5:95-100, 1964

Theory of the Performance Characteristics of Radioisotope Distribution Imaging Systems[1,2]

B. Cassen, Ph.D.

Los Angeles

The current intense revival of interest in the clinical and biological determination of spatial distributions of gamma emitting isotopes and especially in imaging these distributions, despite the crude results commonly obtained, is reflected in the fact that approximately one quarter of the 100 or so papers read at the 1963 Montreal meeting of the Society of Nuclear Medicine deal explicitly with scanning or other imaging techniques. Most imaging of radioisotope distributions is now done with mechanical scanners. Some developments have been undertaken principally by Anger (1) to eliminate the mechanical scanning motions by simultaneously observing the whole field of view. This ingenious principle has advantages and, in its present form, some severe disadvantages as will be analyzed below. In all currently available imaging systems, limited to conventional administered doses of usually used radioisotopes, the images obtained are really very crude. They give some useful information in thyroid scanning and even more marginally in brain tumor, liver, spleen, kidney, pancreas and other partially explored applications. An appreciable increase in overall performance might easily considerably increase the diagnostic value of these procedures. In the following an attempt is made to evaluate an imaging device and an imaging procedure in a quantitative manner so that the factors leading to improvement can be recognized and existing devices and procedures can be compared with each other as to relative merit.

The present discussion is mostly limited to the use of the usual gamma emitting isotopes and there is only casual discussion of the use of annihilation radiation from positron emitters.

The following symbols will be used:

D - dose density, microcuries per cm^3

V - effective resolution volume, cm^3

K - count rate per unit volume per unit dose density per unit solid angle (counts per sec per microcurie per steradian)

W - effective solid angle subtended by detectors to resolution volume, steradians

---

[1]From the Laboratory of Nuclear Medicine and Radiation Biology of the Department of Biophysics and Nuclear Medicine, School of Medicine, University of California at Los Angeles.

[2]These studies were supported by Contract AT(04-1)GEN-12 between the Atomic Energy Commission and the University of California.

F - number of resolution elements in field of view
n - number of resolution elements simultaneously detected
T - time taken to scan or build up image
N - total net counts per source resolution element during the time of scan or build-up
B - total background counts during this time
σ - standard deviation of N
m - figure of merit of system and procedure

The effective resolution volume is in many cases hard to a priori precisely define. However, for practical purposes it can always be semiempirically determined for a specific radioisotope experimentally as follows:

Two equal effective point sources of the radioisotope of interest that give individually counting rates high enough to overwhelm any background counts present are imaged or scanned in air when the sources are in a given plane (or surface) of interest and separated by various preselected distances. The minimum distance apart where the image or scan just shows the presence of two separately resolvable sources is the radius r of an arbitrarily defined spherical resolution Volume $V = \frac{4}{3}\pi r^3$, even though the experimentally observed resolution of the sources placed in a parallel plane a distance of several or many times r away is usually not very different. Perhaps, a more satisfactory definition of V could be evolved but the above is at least based on a well-defined procedure. In an appendix an easy way of preparing point sources is described.

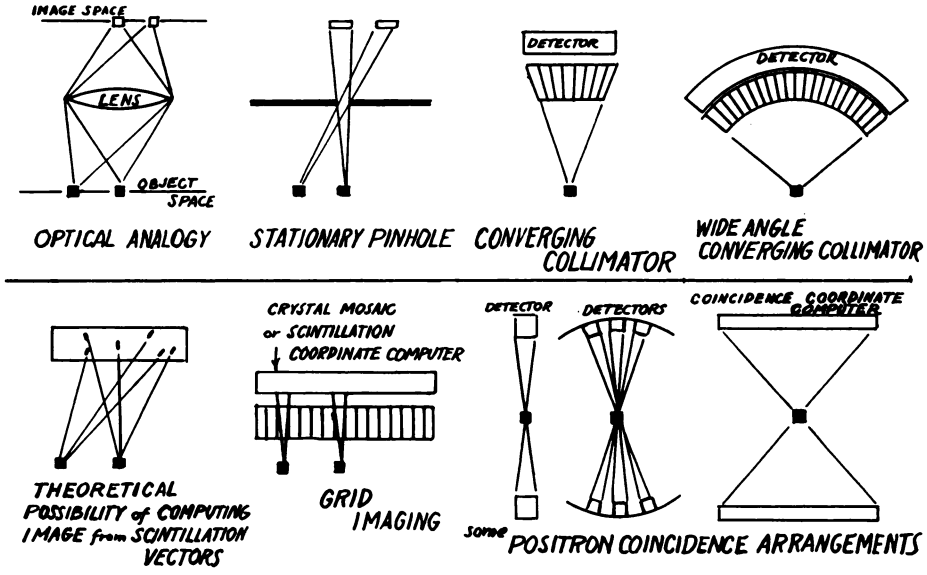

Figure 1

Of fundamental interest is K, the count rate per unit volume per unit dose density per unit solid angle. To measure K for a given isotope one of the point sources can be used if it is calibrated in microcuries. The point source can be considered to be at the center of a resolution volume. Somewhat more accurately a layer of solution of known concentration and thickness 2 r could be used, as the sensitivity to the point source will vary as it is placed in different parts of the resolution volume. However, to make the procedure as simple as possible we will arbitrarily use the point source, although this will make K a little too high in some situations. The solid angle accepted by the detector to the resolution volume is a geometrically determinable factor from the design of the imaging device. It is small for a pinhole or single collimator and can become larger for a converging collimator. In the latter it is important not to count the solid angle subtended by the septa of the collimator.

If we know or measure V, K and W, the net count rate from a resolution volume = K•W•V•D. K•W is the count rate per unit volume per unit dose density and V•D is the number of microcuries in the resolution volume. If the point source is counted, then according to our approximation the effective dose density D during the measurement is microcuries of source / V, so that the experimentally measured count rate = K•W source microcuries, or K•W is directly measurable as count rate divided by source microcuries without knowing W separately. We will not combine K•W to a single factor in order that we can see how changing W can affect performance.

During a scan or image build-up taking a total time T to complete, the time allotted to accumulating counts from a given resolution volume is T / (F/n) = Tn/F. F is the total number of resolution elements in the whole field of view. n is the number of resolution elements measured simultaneously. For example, for ordinary scanning n = 1, a double scanning head with two separate channels would have n = 2. In the extreme, if the whole field of view is detected simultaneously as in the Anger camera, n = F and F/n = 1.

Thereby, the total net counts per resolution element during the complete time T of the imaging process is

$$N = K \cdot W \cdot V \cdot D \cdot T \cdot n \,/\, F$$

Usually, at least in clinical medicine, N is small enough that its standard deviation, arising from Poisson counting statistics, is an appreciable fraction of N. This situation is frequently aggravated by a relatively large background count B caused by activity away from the resolution volume. As $\sigma^2 N = \sigma^2(N+B) + \sigma^2 B = N + 2B$ it follows that

$$\frac{N^2}{\sigma^2} = N \,/\, (1 + 2B/N) = N \,/\, (1 + 2R)$$

where R is the signal to background ratio N/B. Then, a figure of merit can be defined

$$m = \frac{N^2}{\sigma^2} = \frac{K \cdot W \cdot V \cdot D \cdot T \cdot n}{F \cdot (1 + 2/R)}$$

The quality of the image will improve when m increases. When D varies with position, the ability to recognize features of this distribution is contributed to by the images of many neighboring resolution elements. The value of m required for adeqaute recognition of the distribution pattern thereby depends on what sort of information is desired and what might be expected. After m reaches a certain value in critical parts of the image nothing is gained by increasing m further. In many cases recognition is possible with N/σ as low as two, or $m = 4$. Much better results are obtainable with m about 10 and very little is gained in most cases by going much higher.

The factor $K \cdot W \cdot V \cdot n/F$ of m is instrumental, the remaining factor $\frac{D \cdot T}{(1+2/R)}$ arises from procedure. D and T are usually limited by biological or patient considerations. F is sometimes partially adjustable. F must be large enough to cover the field of view of interest with a sufficiently small V to resolve structure of interest, R is partially controllable by proper shielding, although in many cases a part of it comes from activity above or below the resolution plane of interest or from an appreciable penumbra zone of the collimating channels.

In designing an optimum imaging device for a specific application, V and F are fairly well predetermined by the resolution required and the size of the

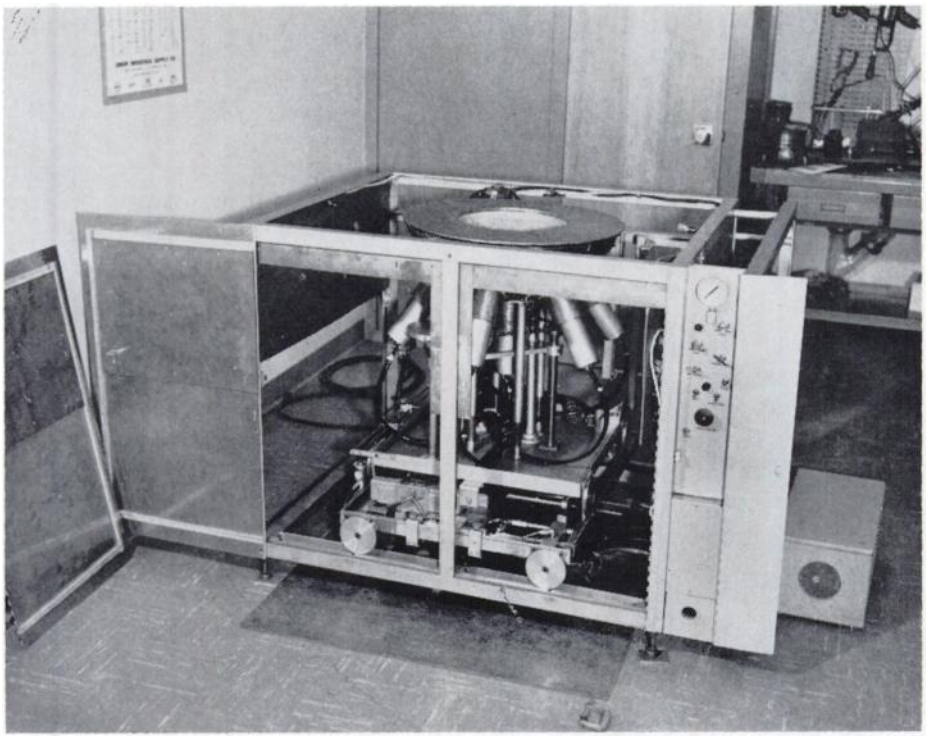

Fig. 2 Appearance of hydraulically driven, 2000 hole collimator scanner during a phase of its construction. The top surface upon which the patient lies is removed. The patient is scanned from below.

field of view required. To make the instrumental contribution to m as large as possible it is necessary to attempt to maximize K•W•n. With a well designed sodium iodide scintillation detector in which the crystal absorbs most of the incident gamma quanta emitted by the radioisotope used, K can not be radically improved. In the extreme of simple one channel scanning with a converging collimator n = 1 and improvement in m can only be made by increasing W. In the other extreme, the Anger camera, n is large and effectively equal to F. Unfortunately, W is small either with a pinhole camera or a grid camera, and for many radioisotopes K is small, as a thin crystal has been used and many quanta go through it without being absorbed to produce scintillations. Bender has improved K by using a mosaic of thicker crystals and using the Anger scintillation coordinate determining system. For a large field of view with good resolution, this system would require a rather large number of mosaic elements.

There are many as yet untried intermediate design possibilities between the extremes n = 1, W as large as possible and n = F, W small. Mechanically scanning a field, especially a large field, with n detectors simultaneously would increase the figure of merit in proportion to n. The arrangement of n detectors with n up to 10 is not too difficult if W the solid angle subtended by each detector is small enough. If W is larger, the size of the single detector becomes large enough so that there is no room for placing several detectors without interfering with each other. For coincidence positron annhilation detection no bulky shielding is required for collimation so that in this case there would be room for more simultaneously scanning channels. Also, in this case, the extra space could alternatively be used to increase the solid angle of acceptance of annihilation coincidences. As, at the present time, there are many more non-positron emitting radioisotopes of clinical interest in imaging than positron emitters, we will limit our discussions to the former, without any implication that there are not interesting and useful developments in positron cameras.

Of equal weight to the solid angle W and the number n of resolution elements simultaneously detected as the dose density D or its product K•D with the sensitivity factor. The dose density can be limited by the specific activity of the administered dose but is more frequently limited on patients by a rather ill-defined consideration of maximum allowable patient radiation dose. Whatever these criteria might be, a non-nuclear beta emmitting isotope, if available, will frequently enable obtaining a large increase in K•D, *e.g.*, substitution of Hg^{197} for Hg^{203}. Also, if there is a choice of an equally useful radioisotope of much shorter half life than the biological half-life, a much larger administered dose can be given for suitably estimated total patient radiation dose. The potentialities of increasing the figure of merit, or taking advantage by decreasing T or V or increasing F and/or N/B, by the use of large doses of short lived isotopes are very considerable. However, being close to a reactor or accelerator or having suitable "cows" to milk off short lived isotopes introduces many practical complications, not the least of which is the rapid chemical processing, sterilization and elimination of pyrogenicity hazards.

Figure 1 diagrammatically illustrates basic types of imaging systems as de-

pendent on W and n. The approximate optical analogy of lens speed and field of view is indicated.

We are now exploring what can be accomplished in improving the figure of merit of a simple scanner by greatly increasing W as diagrammatically indicated in Figure 1 as a wide angle converging collimator. A full description will be given after a clinical testing program has been pursued. The collimator has 2200 holes, subtends a solid angle at the focus of a little less than pi, of which a little more than half is subtended by the collimating channels. The detector consists of about 50 lbs. of irregular pieces of activated sodium iodide immersed in chlorinated diphenyl liquid. The crystals are looked at by seven two-inch photomultiplier tubes. The collimator bowl is hydraulically scanned under the patient. Figure 2 shows the essential appearance of this scanner during a phase of its construction.

SIMPLE METHOD OF PREPARING POINT SOURCES

With a paper punch, making a round hole about 1.5 mm diameter, punch out disk of blotting paper. Place disks in a counting planchet. Drop solution of isotope (approximately twice activity desired in finished source) on disks slowly until saturated. If part of sample remains, dry disks under an infrared lamp and repeat until all of sample is absorbed. Dry disks completely.

Apply small amount of Duco cement to end of small wooden rod (such as Q-tip), pick up 1 disk of blotting paper, let cement dry, repeating until all disks are cemented to end of stick and dried. Coat end of stick with several coats of spray lacquer. Dry thoroughly.

Count completed point source. Then count planchet, container and anything used to apply cement. Obtain activity in point source from original amount used and percentage in point source. Or calibrate point source directly.

If source is too high, part of it can be sliced off with sharp razor blade. Then re-lacquer and recalibrate.

REFERENCE

1. ANGER, HAL O.: Scintillation Camera, *Rev. Sci. Inst.* **29**:27, 1958.