RECEIVED BY DTIC OCT 14 1968

O. R.

Wesbaden Nuclear Medicine
Meetings 9/26-28/68

1000-82

ACRH--1000-82

CONF-680933--1

MASTER

COMPARISON OF GALLIUM-68, TECHNETIUM-99m AND INDIUM-113m USED WITH
THE GAMMA CAMERA AND THE 3-INCH AND 5-INCH SCANNERS FOR
VISUALIZATION OF LESIONS IN THE BRAIN, KIDNEY
LIVER, AND LUNG

K. A. Lathrop, T. D. Cohen, R. N. Beck and P. V. Harper

The Argonne Cancer Research Hospital (operated by the University of
Chicago for the United States Atomic Energy Commission), and
the Departments of Radiology and Surgery, the University
of Chicago, Chicago, Illinois U.S.A.

LEGAL NOTICE

This report was prepared as an account of Government sponsored work. Neither the United States, nor the Commission, nor any person acting on behalf of the Commission:

A. Makes any warranty or representation, expressed or implied, with respect to the accuracy, completeness, or usefulness of the information contained in this report, or that the use of any information, apparatus, method, or process disclosed in this report may not infringe privately owned rights; or

B. Assumes any liabilities with respect to the use of, or for damages resulting from the use of any information, apparatus, method, or process disclosed in this report.

As used in the above, "person acting on behalf of the Commission" includes any employee or contractor of the Commission, or employee of such contractor, to the extent that such employee or contractor of the Commission, or employee of such contractor prepares, disseminates, or provides access to, any information pursuant to his employment or contract with the Commission, or his employment with such contractor.

DISTRIBUTION OF THIS DOCUMENT IS UNLIMITED
Pot

DISCLAIMER

This report was prepared as an account of work sponsored by an agency of the United States Government. Neither the United States Government nor any agency Thereof, nor any of their employees, makes any warranty, express or implied, or assumes any legal liability or responsibility for the accuracy, completeness, or usefulness of any information, apparatus, product, or process disclosed, or represents that its use would not infringe privately owned rights. Reference herein to any specific commercial product, process, or service by trade name, trademark, manufacturer, or otherwise does not necessarily constitute or imply its endorsement, recommendation, or favoring by the United States Government or any agency thereof. The views and opinions of authors expressed herein do not necessarily state or reflect those of the United States Government or any agency thereof.

DISCLAIMER

Portions of this document may be illegible in electronic image products. Images are produced from the best available original document.

INTRODUCTION

When organs are visualized by means of internal radionuclides, maximum information is sought for the least radiation dose to the patient and the shortest examination time. Knowing when this has been achieved has been made more difficult by recent developments in nuclear medicine. One of these developments is the easy availability of many radionuclides all readily modified for selected organ localization and having gamma energies suitable for external detection, while depositing a small radiation dose in tissue either because of a short physical half-life, little nonpenetrating radiation, a short biological half-time, or a combination of these properties. A second development is the commercial production of a variety of imaging systems including single and multiple head scanners, the gamma camera, spark chamber, and image intensifiers. It is difficult to determine the combination of radionuclidic preparation and imaging system with the best detection capabilities for a designated lesion in a particular location of a specified organ for a given radiation dose. Parallel studies in the same patient are limited in applicability. Phantom studies, although more widely applicable, as generally used to simulate the clinical situation are still based on subjective interpretation of images. An objective criterion, the Figure of Merit of Dewey and Sinclair (5), Beck (2,3), and others, has been used for comparison of the physical parameters of scanning systems. This concept may be extended to include the biological distribution and dosimetry (8,10) of radiopharmaceutical preparations. In the present study such considerations have been used in a variety of simulated clinical scanning situations to compute Figures of Merit normalized to one rad absorbed radiation dose to the critical organ. The method itself is general in its application to any radionuclide used with any imaging system. For

illustration, preparations of gallium-68, technetium-99m and indium-113m were used with the Nuclear-Chicago gamma camera and the 3-inch and 5-inch Picker Magnascanners for the detection of lesions in the brain, kidney, liver and lung.

Physical properties of these three radionuclides are summarized in Table 1. ^{99m}Tc has some obvious advantages over ^{113m}In and ^{68}Ga , e.g., radiation shielding of multi-millicurie quantities may be accomplished with lighter weight and less bulky devices; contamination by the parent radionuclide is less serious due to its shorter physical half-life; multiple doses for use throughout the day may be prepared at one time with the relatively longer-lived ^{99m}Tc . The shorter physical half-lives of ^{68}Ga and ^{113m}In do not lower the radiation dosage to the patient since the decay of each of these produces substantial amounts of locally absorbed radiation. However, deep lesions might be detected more efficiently by the more penetrating gamma radiations of ^{68}Ga and ^{113m}In . ^{68}Ga might be used for determining the depth location of a lesion by taking advantage of the focusing capability of the positron attachment for the gamma camera. In addition, it might be possible in some instances to achieve a more favorable biological distribution by using indium or gallium rather than technetium.

The Figure of Merit embodies reciprocal relationships between patient radiation dosage, scanning time, resolution, and statistical reliability for detection of a lesion. Although statistical detection of a lesion and its perception by the physician examining a scan are not identical, it is evident that they are related, and mathematical models of perception similar to that used for detection have been proposed (11,12). Clearly, anything perceptible is detectable, but the reverse may not be true since

a statistically detectable object may not be perceived by the eye-brain system. A high value for the Figure of Merit indicates a high rate of build-up for the signal to noise ratio in the region of interest and consequently a more rapid achievement of perceptibility than does a low value.

The Figure of Merit, as used in the present study, includes consideration of the biological distribution and radiation characteristics of the radiopharmaceutical preparation, the detection capabilities of the imaging system, and the size and depth of the lesion. Once determined, the biological measurements for a radiopharmaceutical may be used to calculate Figures of Merit for other imaging systems, and similarly, physical measurements for one radionuclide may be applied to other biological systems. Biological distribution data in humans are usually very difficult or impossible to obtain, necessitating reliance on data from laboratory animals for estimation of the human situation. Since it is also difficult to measure organ to tumor ratios of radionuclide concentration, the use of values obtained with realistic phantoms presents a reasonable solution for determining the object contrast for purposes of comparing radionuclides. The emitted gamma flux over an organ, however, is easily determined in a human subject with a detector of known sensitivity; and for comparative studies, relative organ count rates per injected millicurie suffice. Measurements involving the detection capabilities of the imaging system are the modulation transfer function and sensitivity, which, once available, may be applied with the biological data outlined above for any pharmaceutical preparation of the designated radionuclide. For example, the absorbed radiation dose for ^{99m}Tc -S colloid

calculated from biological distribution data, and the effective object modulation and emitted gamma flux for the specified lesion apply for any imaging system. Similarly, the MTF and sensitivity data for ^{99m}Tc obtained with, for example, the gamma camera, may be used for any ^{99m}Tc pharmaceutical.

METHODS

Calculation of the Figure of Merit (Q). This formulation takes into account the complete system involved in the scanning situation (3) which consists of the patient whose physiological processes have distributed the radionuclide, and the collimator and detector with its associated electronics which collimate and produce pulses from the photons emanating from the patient. The situation in the patient is thought of as a lesion of specified size embedded in normal organ tissue. If C_t and C_o are the true mean count rates over the lesion and normal tissue respectively, the difference $(C_t - C_o)$ is regarded as the signal to be detected. The noise accompanying this signal is simply the random fluctuations in the signal itself, which can be measured by the standard deviation of the signal;

S.D. $(C_t - C_o) = \sqrt{\frac{C_t + C_o}{\tau}}$, where τ is the observation time for C_t and for C_o . The criterion for detection is that the signal-to-noise ratio be equal to or greater than some assigned value;

$$\text{SNR} = \frac{\text{SIGNAL}}{\text{NOISE}} = \frac{(C_t - C_o)}{\sqrt{\frac{(C_t + C_o)}{\tau}}} \equiv 2z, \quad (3)$$

where z = the number of standard deviations, which is multiplied by 2 since both false positive and false negative values must be considered.

The observation time required to accumulate enough counts to resolve C_t and C_o in this situation is

$$\tau \geq \frac{4z^2 (C_t + C_o)}{(C_t - C_o)^2} = \frac{4z^2}{Q}$$

This equation defines Q as

$$Q = \frac{(C_t - C_o)^2}{(C_t + C_o)} = \frac{(\text{SNR})^2}{\tau} = \frac{4z^2}{\tau}$$

which can be interpreted as the rate at which the (signal-to-noise ratio)² is built up by the imaging system. If C_t and C_o are the maximum and minimum count rates in response to a sinusoidal plane wave of radioactivity with frequency ν , then $Q(\nu)$ can be interpreted as a measure of the detectability of such a sinusoid. In that case, $Q(\nu)$ can be expressed in terms of the average count rate $\bar{C}(\text{cpm}) = (C_t + C_o)/2$ and the image

$$\text{modulation } m_I(\nu) = \frac{(C_t - C_o)}{(C_t + C_o)},$$

$$Q(\nu) = 2\bar{C}m_I^2(\nu)$$

Since the image modulation is given by the product of the effective object modulation $m_o/(1+S)$ and the effective modulation transfer function $\text{MTF}_G/(1+P)$, we can also write

$$Q(\nu) = 2\bar{C} \left[\frac{m_o(\nu)}{1+S} \times \frac{\text{MTF}_G(\nu)}{1+P} \right]^2$$

where m_o = the object modulation

S = scatter fraction

MTF_G = modulation transfer function when septal penetration is negligible

P = the penetration fraction (3).

If the radiation dose to the region of greatest exposure is d (rads/mCi injected) and the average count rate is $\bar{C}$ (cpm/mCi injected), the Figure of Merit per rad of exposure is

$$\frac{Q(v)}{d} \left[\frac{\text{cpm}}{\text{rad}} \right] = \frac{\bar{C}}{d} \left[\frac{m_o(v)}{1+S} \times \frac{MTF_G(v)}{1+P} \right]^2$$

Quantities that must be measured are the line spread function, effective object modulation, biological distribution with time, and organ counts per minute per cm^2 per injected mCi. Each of these measurements is sensitive to the pulse height analyzer window, which was set at twice the full width at half-maximum except for the gamma camera, where a 20% window with ratio circuitry was used.

Line spread functions were measured according to the method specified by the IAEA (7) at appropriate distances from the collimator face. In addition, the ^{68}Ga source was enclosed by steel tubing 2.5 mm in diameter with one mm walls to convert the positrons to the annihilation radiation. The modulation transfer functions (Fourier transform of the line spread function) were calculated using the computer program written by Sylvia Fedoruk (4). From these curves, MTF values for calculating Figures of Merit were chosen by assuming that the effective lesion size was approximated by a plane wave whose wavelength is twice the lesion diameter ("half-wavelength approximation"). For 1-inch lesions the wavelength is thus 0.2 cycles per cm, and for 2-inch lesions is 0.1 cycles per cm. Since the line spread measurements were made for each nuclide with the collimator selected for the various situations, the effect of septal penetration is included. The determined value is actually the effective MTF, equal to $MTF_G/(1+P)$. The focal planes of the collimators are assumed to be two inches below the

body surface, the collimator faces of the camera in contact with the body surface, and the focal plane of the coincidence detector at the level of the lesion.

Effective object modulation is computed from the target to nontarget count ratio, where the target may be either a hot or a cold lesion surrounded by radioactivity. In all cases the lesion was large enough to cover completely the field of view of the collimator, a very fine resolution single hole with an MTF_G nearly equal to one and with no septum penetration. Values for the effective object modulation were determined by use of realistic phantoms as specified for each organ in Table 2.

The phantom measurements take account of the effects of isotope distribution, and absorption and scatter in the surrounding tissue but not the effect of lesion diameter which is determined by the collimator MTF at the appropriate distance from the collimator face. All lesions are assumed to be cylindrical, one inch in diameter by one inch in thickness, except for the lung where they are two inches by two inches.

Using these systems, measurements include radiation scattered by the surrounding medium so that the calculated modulation contains the scatter fraction. By definition, when the $MTF = 1$, the effective object modulation, $m_o / (1 + S)$, is equal to the image modulation; i.e., the difference of the count rate over a lesion and the count rate over an equal area of normal tissue divided by the sum of these two. The modulation value for the annihilation radiation coincidence system of the gamma camera was computed from the attenuation coefficients for the 511 keV gamma in tissue, since the effects of scatter and attenuation in this situation are slight. For these measurements, the detector system used with ^{99m}Tc and ^{113m}In consisted of a collimator made by drilling a one-quarter inch hole through a 2-inch lead brick, which was backed by a 5-inch NaI crystal. For ^{68}Ga

measurements, the collimator was a one-half inch hole drilled through 4 inches of lead, with additional lead surrounding the crystal to eliminate penetration of the 511 keV gamma.

Biological distributions of radiopharmaceuticals were determined in mice at intervals found appropriate for the biological half-time of the preparation and the physical half-life of the radionuclide. Biological half-times in humans were compared with those in mice by measuring radioactivity in the excreta. The radiopharmaceuticals used are summarized in Table 3. Many of these are well established clinical agents whose preparation and distributions may be found in the literature. Others were developed in our laboratories and are not to be considered as the optimum preparation that might ultimately be produced. A primary purpose of this entire investigation was to determine which radionuclide might most advantageously be developed for a given scanning situation. Indium-113m could be modified directly in the eluate (0.05 N HCl, pH 1.3). The ^{113m}In used in this investigation was eluted from a New England Nuclear generator in which ^{113}Sn is held on an ion-exchange bed of hydrous zirconium oxide. The nature of some of the preparations described probably depends on small amounts of the column material contained in the eluate, ^{113m}In eluates obtained from generators prepared with other ion-exchangers might behave differently. It was necessary to process the gallium-68 eluate (0.005 M $\text{Na}_3\text{-EDTA}$) by extracting it as gallium chloride into ethyl ether, evaporating off the ether and dissolving the ^{68}Ga into the desired reagent. Using the biological distribution parameters established for each preparation, the absorbed dose in rads to the regions of principal concentration was calculated by conventional dosimetry using the assumptions

shown in Table 4. Doses computed by the newer method of absorbed fractions (9) would not differ significantly from these for gammas in this energy range. The doses most pertinent to this investigation have been summarized in Table 5.

Organ counts per minute per cm^2 per injected millicurie over the region of interest were measured on human subjects by using the gamma camera together with a 1600 channel analyzer after injection of a carefully calibrated radionuclide preparation. Measurements over the kidney were corrected for body background by subtraction of counts collected over a similar area of the abdomen without kidney. Such a correction was necessary only for the kidney where the circulating radioactivity remained high during the first 12 hours after injection. These corrections were applied to the object contrast determined in the phantom. The collected counts were integrated over a uniform area within the borders of the region scanned. The number of channels per cm was determined by measuring a line source in two positions 10 cm apart. This measurement determines, for the camera configuration, the counts per minute per cm^2 per mCi or, with the dosimetric computation of rads to the critical organ per mCi, the counts per minute per cm^2 per rad. Using the relative sensitivity measurements between the scanner and camera this result can be transformed to the scanner, or, if desired, the reverse procedure may be carried out. In the case of coincidence measurements of positron annihilation using the camera, the maximum total count rate capability of about 10,000 counts per minute imposed substantial additional limitations in some cases. The relative sensitivities were determined using an infinite (12-inch diameter) plane source of the radionuclide. Since the camera has a useful area of detection

of 507 cm² (10-inch diameter), it was assumed for purposes of comparison that the scanner views an equal area.

Characteristics of the collimators used in making these measurements are shown in Table 6. All of these collimators, both for the camera and scanners, are more finely focused than is required for optimal detection of the fundamental object wavelength. This is particularly true of the collimators used with ^{99m}Tc with the scanners that sacrifice sensitivity in favor of resolution to a marked extent. Substantial septum penetration occurs with both ⁶⁸Ga and ^{113m}In for the collimators available with the 3-inch scanner. Results from measurements with this equipment thus represent practical observations rather than optimal comparisons. Effects of using optimal collimation will be considered in the discussion.

RESULTS AND DISCUSSION

The value determined for the Figure of Merit in each of the specified situations is shown in Table 7. In all cases except one the Figure of Merit for the gamma camera is higher than for either scanner. The exception, ⁶⁸Ga in the lung with the coincidence detector system, is explained by the count rate limitation of the annihilation radiation detector, which is unable to make full use of the emitted gamma photon flux. This same consideration also lowers the value of Q/rad for ⁶⁸Ga in the liver using the coincidence detector, although to a much less extent, because of the greater tissue absorption. This problem of count rate limitation is not present with small areas of radioactivity, as shown by the high Q/rad values of ⁶⁸Ga in the brain and kidney, the latter being the only instance where ^{99m}Tc does not have the highest number. Positron annihilation radiation seems useable in this situation particularly where the location of the lesion can be

reasonably well estimated, as in the kidney, so that its tomographic capability can be fully utilized. In the brain, this type of detection could also be exploited for the tomographic capabilities shown by the equally good values for surface and deep lesions. The thin NaI crystal (1/2 inch) of the gamma camera is highly efficient for the 140 keV gamma from ^{99m}Tc , but not for the 393 keV energy from ^{113m}In which results in a lower Q/rad value.

The very high Figures of Merit for the lung represent exceedingly short detection times, a few milliseconds for 99.9% confidence with ^{99m}Tc , and suggest the superiority of this radionuclide for dynamic perfusion studies. Although values for ^{99m}Tc are higher than those for ^{113m}In in the lung, it is much easier to make a preparation of ^{113m}In carried on an inert hydroxide than macroaggregates of high specific activity Tc-labeled albumin. For this reason, many laboratories prefer to use the former. Perhaps the preferable alternative is perfection of a more easily prepared ^{99m}Tc lung-scanning pharmaceutical.

As anticipated, the 5-inch scanner with its NaI crystal of greater diameter than that of the 3-inch scanner (and consequently a longer collimator), is more efficient for ^{113m}In , and higher Q/rad values are obtained with this instrument than with the 3-inch scanner. With ^{68}Ga , septal penetration is still quite high with the collimator for the 5-inch crystal, and the same increase in Q/rad is not realized. ^{99m}Tc and ^{113m}In have comparable values when the 5-inch scanner is used. With both rectilinear scanners, Figures of Merit are lowered for ^{68}Ga and ^{113m}In because the high septal penetration degrades the modulation transfer function even when using the best collimators available for this study. Line spread functions determined for ^{68}Ga were measured with the source encased in a

steel tube to convert the positrons at their source. In tissue, the positrons travel farther before being annihilated, which would give greater degradation of the resolution than is evident from the experimental studies, and consequently Q/rad for this radionuclide would be lowered in actual practice.

It is interesting to consider the effect that optimal collimation would have on the results presented here. It has been shown (3) that optimal geometrical collimation for a given space frequency involves a diameter of view approximately equal to the wavelength, or a full-width-at-half-maximum equal to the lesion diameter (half-wavelength approximation). Even the coarse focus collimators used here have a resolution twice as fine as this optimal value. This very coarse focus optimal collimation is of course optimal only with respect to detection, and depicts lesions very poorly. Nevertheless, it is instructive to look at the effect such an "optimal" collimator would have on the Figures of Merit. The $^{99\text{m}}\text{Tc}$ collimators used are comparatively fine focus, since the 50% isoresponse region is one-fourth inch for the 3-inch scanner, and three-sixteenths inch for the 5-inch scanner, as compared to a 50% isoresponse region of one-half inch for the collimators used with ^{68}Ga and $^{113\text{m}}\text{In}$. They are thus considerably less optimal than the $^{113\text{m}}\text{In}$ and ^{68}Ga collimators, so that optimal geometrical collimation would increase the $^{99\text{m}}\text{Tc}$ Figures of Merit substantially relative to those for ^{68}Ga and $^{113\text{m}}\text{In}$, especially for the 5-inch scanner. The magnitude of this effect has been estimated (3) for cases lying at the focal distance. The Figure of Merit increase for the high energy collimators is by a factor of 4, while for the low energy collimators the increase is by a factor of 7.5 for the 3-inch scanner and

10 for the 5-inch scanner. The effect of using optimal geometric collimation is thus a relative improvement in the ^{99m}Tc Figures of Merit by a factor of 2 to 2-1/2 for the scanners. For the camera at 3 inches from the collimator face, optimization of the collimators improves the Figures of Merit by approximately a factor of 4 for both ^{113m}In and ^{99m}Tc . The question of collimator optimization does not apply to the coincidence system, which does not achieve improved resolution by excluding emitted photons. Thickening of the septa of the high energy collimators for the scanners used with ^{68}Ga and ^{113m}In would reduce septal penetration and hence increase the image contrast, but the sensitivity would be reduced. The overall gain in Q/rad achievable by this process is probably less than a factor of two.

When Q/rad for a radiopharmaceutical is known for one imaging system and object (A), the corresponding value for some other system (B) can be obtained readily after determining $\text{MTF}_{(B)}$ and $\text{Sensitivity}_{(B)}$

$$Q/\text{rad}_{(A)} \times \left[\frac{\text{MTF}_{(B)}}{\text{MTF}_{(A)}} \right]^2 \times \frac{\text{SENSITIVITY}_{(B)}}{\text{SENSITIVITY}_{(A)}} = Q/\text{rad}_{(B)}$$

Similarly, when Q/rad for any imaging system is known for one radiopharmaceutical (D), the Q/rad for another pharmaceutical (E) of the same radionuclide used with the same imaging system for the same object may be obtained after determination of the dose to the region of greatest exposure and the cpm/mCi from the region of interest by,

$$Q/\text{rad}_{(D)} \times \frac{\frac{\bar{C}}{d} (E)}{\frac{\bar{C}}{d} (D)} = Q/\text{rad}_{(E)}$$

Inherent in this application of the Figures of Merit is the amount of radioactivity which may be injected to deliver one rad to the critical

organ. For ^{113m}In and ^{99m}Tc these quantities are very comparable, the greatest variation being less than a factor of three greater for ^{113m}In in the kidney. Costs per patient dose are comparable for each of the preparations, but can be determined accurately only in terms of conditions existing within the particular clinical situation. It is our belief that when a choice exists, evaluation of a radionuclide preparation must be made by each nuclear medicine installation in terms of available imaging equipment, patient dosage, clinical volume, radiation exposure to personnel, and available technical capabilities. As far as the actual procedure of comparison is concerned, the method used here appears to encompass most of the accessible parameters.

SUMMARY

A general method is presented for evaluating the combination of a radiopharmaceutical and an imaging system by which data can be applied interchangeably to other combinations for calculating Figures of Merit for detectability. The method is illustrated by use of preparations of ^{68}Ga , ^{99m}Tc and ^{113m}In with the Nuclear Chicago Gamma Camera, the 3-inch Picker Magnascanner, and the 5-inch Picker Magnascanner for visualization of the brain, kidney, liver and lung. In general, the highest Figures of Merit were obtained with ^{99m}Tc . Except in the case of the kidney and brain, ^{68}Ga is severely limited when imaged with the annihilation radiation detector of the gamma camera because of the count rate limitation inherent in this system. ^{113m}In performs better with the 5-inch scanner than with the 3-inch scanner.

REFERENCES

1. Anger, Hal O. and Gottschalk, Alexander. Localization of Brain Tumors with the Positron Scintillation Camera. J. Nucl. Med. 4: 326-330 (1963).
2. Beck, Robert N. A Theoretical Evaluation of Brain Scanning Systems. J. Nucl. Med. 2: 314-324 (1961).
3. Beck, Robert N. and Harper, Paul V. Criteria for Evaluating Radioisotope Imaging Systems, Ch. 30. In Fundamental Problems in Scanning (Gottschalk, A. and Beck, R. N., Eds.) Charles C Thomas, Publishers, Springfield, Illinois (1968).
4. Craddock, T. D. Assessing the Performance of Radioisotope Scanners: Data Acquisition. J. Nucl. Med. 9: 210-218 (1968).
5. Dewey, W. C. and Sinclair, W. K. Criteria for Evaluating Collimators Used in In Vitro Distribution Studies with Radioisotopes. Intern. J. Appl. Radiation Isotopes. 18: 815-823 (1967).
6. Harper, Paul V., Lathrop, Katherine A., and Gottschalk, Alexander. Pharmacodynamics of Some Technetium-99m Preparations. In Radioactive Pharmaceuticals, Proceedings of a Symposium held at the Oak Ridge Institute of Nuclear Studies, Nov. 1-4, 1965. Available as CONF-6511111 from Clearing house for Federal Scientific and Technical Information, National Bureau of Standards, U. S. Department of Commerce, Springfield, Virginia, pp. 335-358 (1966).
7. Hine, Gerald. Evaluation of Focused Collimator Performance. II. Digital Recording of Line-Source Response. Intern. J. Appl. Radiation Isotopes. 18: 815-823 (1967).

8. Kuhl, David E. Influence of Collimator Design, Photon Energy and Radiation Dose on Detection Effectiveness in Liver Scanning. *Phys. Med. Biol.* 10: 93-105 (1965).
9. Loevinger, Robert and Berman, Mones. A Schema for Absorbed-Dose Calculations for Biologically-Distributed Radionuclides. *J. Nucl. Med., Suppl.* No. 1, February (1968).
10. Matthews, Christine M. E. Comparison of Isotopes for Scanning. *J. Nucl. Med.* 6: 155-168 (1965).
11. Morgan, Russell H. Threshold Visual Perception and its Relationship to Photon Fluctuation and Sine Wave Response. *Am. J. Roentgenol. Radium Therapy Nucl. Med.* 93 (4): 982-997 (1965).
12. Rose, Albert. The Sensitivity Performance of the Human Eye on an Absolute Scale. *J. Opt. Soc. Am.* 38 (2): 196 (1948).

Complete address of Senior Author:

Katherine A. Lathrop
Argonne Cancer Research Hospital
950 East 59th Street
Chicago, Illinois 60637 U.S.A.

TABLE 1

PHYSICAL PROPERTIES OF THREE SHORT-LIVED GENERATOR ISOTOPES

PROPERTY	Gallium-68	Indium-113m	Technetium-99m
Physical half-life	68 minutes	1.7 hour	6 hours
Parent, half-life	Ge-68, 260 days	Sn-113, 118 days	Mo-99, 67 hours
Principal gamma energy	511 KeV, $.879 \times 2 \gamma/d$	393 KeV, $.67 \gamma/d$	140 KeV, $.92 \gamma/d$
Specific gamma constant (Γ)	5.04	1.45	0.56
$\bar{E}$ (MeV)(nonpenetrating radiation)	.53 (β^+, ϵ)	.126 (ϵ)	.014 (ϵ)
Half-value layer			
Water	7.3 cm	6.8 cm	4.3 cm
Lead	.4 cm	.3 cm	.02 cm
Sodium Iodide	2.0 cm	2.1 cm	.24 cm

TABLE 2

PHANTOM DIMENSIONS FOR MEASUREMENT OF EFFECTIVE OBJECT (LESION) MODULATION: $m_0/(1 + S)$

Organ		Phantom		Lesion		Body/Organ/Lesion Activity Ratio	Remarks
Thickness	Depth	Diameter	Thickness	Depth in Organ	Thickness		
Brain	8"	-	8"	8"	3-1/2"	1"	1:1:5
Kidney	1"	3"	8"	7-1/2"	-	1"	0:1:0 Correction made for observed body background.
Liver	4"	1"	8"	8"	0",1",3"	1"	0:1:0 Lesion depth measured from surface of organ to sur- face of lesion for liver and lung.
Lung	4"	1"	8"	6"	0",2"	2"	0:1:0 Lesion and organ simulated by air-filled gelatine capsules surrounded by activity in solution.

TABLE 3

Organ	Radiopharmaceutical			Preparation
	Ga-68	Tc-99m	In-113m	
Brain	Ga-EDTA (1)	Na Tc O ₄ (6)	In-EDTA	Equal volumes eluate and 0.005 M Na ₃ EDTA mixed and neutralized to pH 6-7 with NaOH. Sterilized by filtration through 0.22 millimicron membrane filter.
Kidney	Ga-Fe complex	Tc-Fe complex (6)	In-Fe complex	2 ml of 2.5 N HCl solution containing Ga-68 or In-113m mixed with 0.1 ml 5% ascorbic acid and 0.1 ml of 1% FeCl ₃ ·6H ₂ O in 1% HCl neutralized to pH 7.8 with NaOH and sterilized by filtering through 0.22 millimicron membrane.
Liver	Ga-Zr colloid	Tc-S colloid (6)	In-Zr colloid	Dry, extracted Ga-68 dissolved in 0.05 N HCl containing 4 microgram/ml zirconium, neutralized to pH 5.5, using sterile reagents and containers. Sterile In-113m eluate neutralized to pH 5.5 with sterile NaOH.
Lung	Ga-Fe aggregates	Tc-albumin Macro- aggregates (6)	In-Zr aggregates	Dry, extracted Ga-68 dissolved in 0.05 N HCl containing 500 micrograms Fe ⁺⁺⁺ , neutralized to pH 7.5, using sterile reagent and containers. In-113m eluate made basic with NaOH to about pH 11, neutralized to pH 8 with HCl and sterilized by autoclaving.

TABLE 4

DOSIMETRY PARAMETERS AND ASSUMPTIONS (1 mCi)

ORGAN	SHAPE	SIZE	$\bar{g}$	MASS (g)	$\mu\text{Ci/g}$ C_0	T_{eff}	
Kidney	Spheres	R = 3.3 cm	35	150 - Cortex	$\frac{50}{300}$ (Ga-68)	T_p	Kidney agents fixed in cortex (β dose)
					$\frac{150}{300}$ (Tc-99m)		γ dose to whole organ (EDTA)
				300 - Total	$\frac{200}{300}$ (In-113m)		
Liver	Sphere	R = 7.2 cm	59	1500	$\frac{800}{1500}$	T_p	
Lungs	Cylinder	H = 20 cm	67	1000	$\frac{800}{1000}$	T_p	Lungs together as a unit
		R = 10 cm				$\frac{T_p}{2}$ (Tc-99m)	
(Brain) Stomach and Contents	Sphere	R = 4 cm	$g_p = 25$	300	$\frac{150}{300}$ (Ga-68)	T_p (Ga)	β dose = C_0 equal in stomach and contents. γ dose = mucosa at surface of sphere.
					+ In-113m)	T_p (Tc)	
					$\frac{300}{300}$ (Tc-99m)	$\frac{T_p}{2}$ 36 min (In)	

TABLE 5

RADS PER INJECTED MILLICURIE TO ORGAN RECEIVING HIGHEST DOSE

ORGAN VISUALIZED	CRITICAL ORGAN	Ga-68	In-113m	Tc-99m
BRAIN	Ga-68, Kidney In-113m, Kidney Tc-99m, Stomach	1.05	0.14	0.19
Kidney	Kidney	0.67	0.96	0.34
LIVER	Liver	1.23	0.46	0.29
LUNG	Lung	1.56	0.53	0.25

TABLE 6

CHARACTERISTICS OF COLLIMATORS

RADIO- NUCLIDE	Gamma Camera	3-inch Magnascanner					5-inch Magnascanner				
		Colli- mator No.	Number of Holes	Focus	Focal Depth (Inches)	50% Iso- response (Inches)	Colli- mator No.	Number of holes	Focus	Focal Depth (Inches)	50% Iso- response (Inches)
Ga-68	Positron Head	2107	19	Medium	3	$\frac{1}{2}$	2111	85	Coarse	5	$\frac{1}{2}$
Tc-99m	$1\frac{3}{4}$ " 4000 hole	2113	73 (low energy)	Medium	3	$\frac{1}{4}$	2114	265 (low energy)	Fine	3	$\frac{3}{16}$
In-113m	3 " 1000 hole	2107	19	Medium	3	$\frac{1}{2}$	2111	85	Coarse	5	$\frac{1}{2}$

TABLE 7

FIGURES OF MERIT (Q/rad) FOR DETECTING VARIOUSLY LOCATED LESIONS

RADIONUCLIDE	INSTRUMENT	BRAIN	KIDNEY	Liver			Lung	
				Lesion depth [†]			Lesion depth [†]	
				Surface	1-inch	3-inches	Surface	2-inches
Ga-68	Gamma Camera	8.2	230.0	5.3	5.3	5.3	62.0	62.0
	3"-Magna Scanner	.12	2.9	2.5	0.58	0.13	190.0	92.0
	5"-Magna Scanner	.14	3.5	2.3	0.47	.20	130.0	85.0
Tc-99m	Gamma Camera	20.0	88.0	540.0	172.0	42.0	9500.0	4800.0
	3"-Magna Scanner	1.7	7.0	39.0	14.5	2.9	690.0	325.0
	5"-Magna Scanner	1.8	7.5	27.0	16.0	2.2	550.0	310.0
In-113m	Gamma Camera	4.8	15.0	71.0	17.0	8.3	1400.0	600.0
	3"-Magna Scanner	1.3	3.9	17.0	3.6	1.6	200.0	86.0
	5"-Magna Scanner	3.8	11.0	25.0	9.6	5.5	550.0	320.0

[†]Measured from organ surface to top of lesion.