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ORO-2401-87

AN IN-VIVO EVALUATION OF STANDARD MAN MODEL ABSORBED FRACTIONS USING ^{99m}Tc-SULFUR COLLOID

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ABSTRACT

This study was performed to check how well the calculated absorbed fraction data (S-factors) given in MIRD Pamphlet No. 5 and related reports apply to living patients. An external target region was defined on the skin of the Snyder-Fisher phantom, anterior to and overlaying the center of the liver. This target was a 5 cm square by 0.089 cm thick slab of LiF, and the S-factors were computed for this target and uniform distributions of ^{99m}Tc in the liver and spleen by J.W. Poston and G.G. Warner of ORNL. Experimentally, this target was represented by placing thirteen LiF TLD's of the same thickness in a correspondingly positioned 5 cm square area. An experiment performed with the MR. ADAM phantom showed good agreement between the measured (TLD) and calculated (S-factor) doses.

In the patient studies, the group of thirteen TLD's was correspondingly positioned, and the measured TLD dose compared to that calculated from the patient cumulated activity and the Snyder-Fisher phantom S-factors. In most cases, these doses agreed to within 30%, although larger discrepancies were observed with non-standard sized patients. Based upon these results, it is desirable to make further investigations of this type, especially with lower energy gamma emitters, and the methods of dose and activity measurement used in this study should be useful unless the photon energy is too low, or the source activity changes rapidly with time.

INTRODUCTION

The primary motivation for the extensive work in calculating absorbed fractions and S-factors is to give parameters suitable for accurate patient dose calculations. Since the S-factors are computed using a mathematical model, it is important that their applicability to living patients be checked. The direct method for this would be to measure the radiation dose to a patient target region, and compare this to that calculated using the appropriate S-factors. However, patient target regions are either internal organs or total body skin, so direct dose measurements would be difficult at best. Our approach was to define a special target region on the skin of the patient, for which the dose was easily measured with no patient discomfort. The major disadvantage was that a special computer run was needed to compute the S-factors for this target.

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EXPERIMENTAL ARRANGEMENT

The source-target configuration is indicated in Figure 1. The source organs were the liver and spleen, each filled with a uniform distribution of

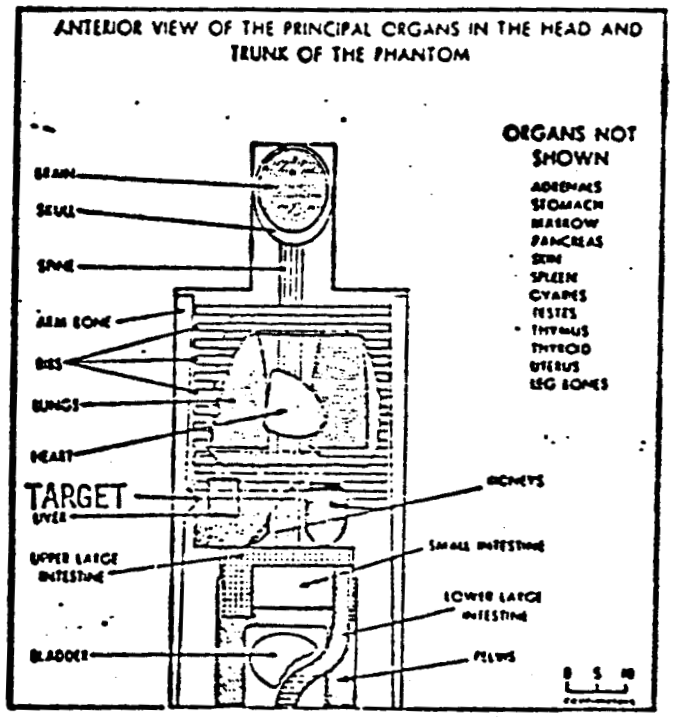

Figure 1. The source-target geometry.

^{99m}Tc. The target was a 5 cm square by 0.089 cm thick slab of LiF, on the skin anterior to the liver as shown. S-factors for the liver to LiF and spleen to LiF configurations were calculated through the kindness of W.S. Snyder, J.W. Poston, and G.G. Warner of ORNL, using their Monte Carlo techniques.

Experimentally, this source distribution was approximated by using patients injected with ^{99m}Tc-sulfur colloid for routine liver/spleen imaging. The target was approximated by placing thirteen LiF TLD's of the same thickness as the slab in a 5 cm square area over the center of the liver. In order to assess the applicability of the S-factors, one must be able to accurately measure the TLD dose and the cumulated activity in the liver and spleen separately. Therefore, the methods of measurement used were carefully checked and evaluated.

METHOD OF ANALYSIS

The cumulated activities were found by integrating the measured activity with respect to time. The activity, in turn, was measured by scanning techniques, with the scanning system schematically outlined in Figure 2. The patient lay on a stationary bed between dual opposing NaI(Tl) detectors, and counts from the 140 keV photopeak of ^{99m}Tc were stored in a PDP-9 computer as the detectors moved in a rectilinear raster motion. In addition, a ^{133}Ba source (356 keV) was placed in the top collimator and shielded so that a narrow photon beam passed down through the patient to the bottom detector. These counts were also recorded, giving the emission (^{99m}Tc) and transmission (^{133}Ba) scans simultaneously.

The method used to compute organ activity from the scan data is outlined

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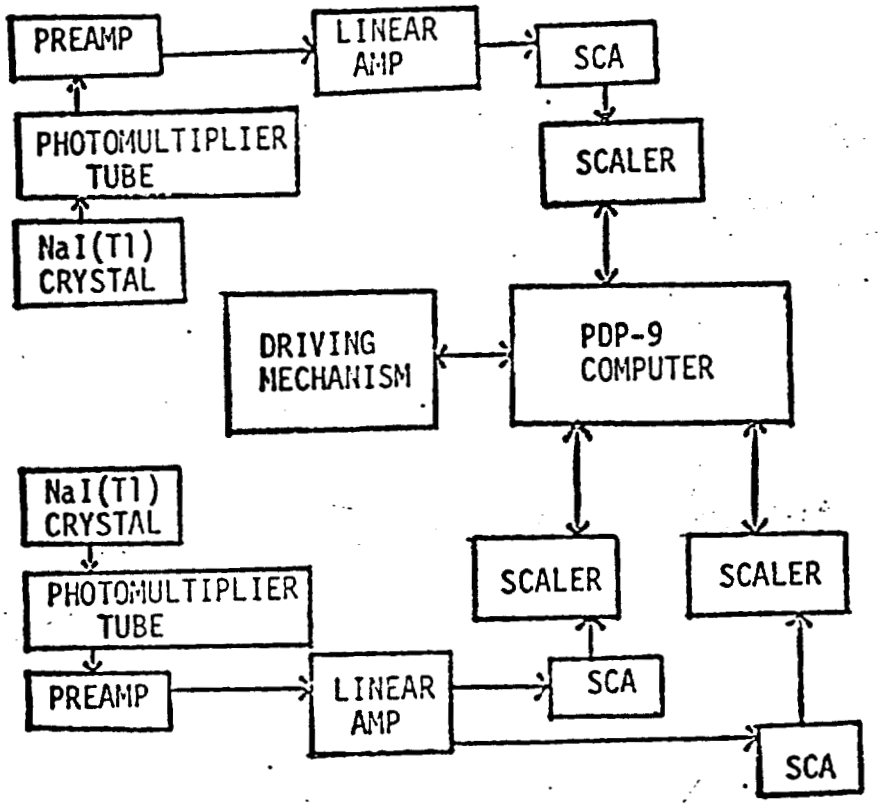

Figure 2. Schematic of scanning system.

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in Table 1. The basic approach was to compare

Table 1. Analysis of scan data

I. For liver activity
   1. Sum top and bottom detector counts individually
   2. Subtract away background sums
   3. Compute the geometric mean (GM) of these sums
   4. Calculate the equivalent thickness d from

$$\exp(-\mu d/2) = m^{-1} \sum_{j}^{m} \exp(-\mu dj/2)$$

   5. Compute $\bar{A} = GM/F(d)$
   6. Compute the activity A from

$$A = \bar{A} \cdot [\sum_{j}^{m} \sinh(\mu tj/2)/\sum_{j}^{m} (\mu tj/2)]^{-1}$$

II. For spleen activity:
   1. Do steps 1-5 above
   2. Assume $A = \bar{A}$.

the patient geometric mean counts to F(d), the geometric mean counts/mCi obtained from scans of a point source within a water phantom of thickness d. First, the equivalent patient thickness d was calculated ((the dj are the set of body thicknesses for a given patient), because the calibration data F(d) had been obtained with a constant-thickness water phantom. From this, a preliminary activity estimate, $\bar{A}$, was found, and in the last step, $\bar{A}$, is corrected for self-attenuation in the liver, giving the activity estimate A. The set of liver thicknesses, tj, were determined from ultrasonic scans. A similar analysis was done for the spleen activity, except that a correction for self-attenuation was not made, since the spleen is rather thin and also contributed relatively little dose to the TLD's. From sequential measurements of the measured activity, the cumulated activity was easily found. It was observed that the decrease of organ activity was due entirely to the physical decay of ^{99m}Tc, as expected. A series of phantom scans was also run to check the accuracy of this technique; the results are shown in Table 2.

The accuracy of the dose inferred from TLD readings was also assessed. By repeated exposures, it was found that the reproducibility of a TLD reading could be kept at approximately ±1% at the 30-100 mrad dose level, provided that one accounted for variations in reader pan reflectivity and electronics drift, and took care to keep these variations small.

An experiment was then run at ORNL with the MR. ADAM phantom to check for possible systematic error in the method of dose measurement. Under the direction of J.W. Poston and P.S. Stansbury, hollow liver and spleen "organs" were fabricated at ORNL. Then TLD measurements were made on MR. ADAM for activity in the liver only, and repeated for activity in the spleen only. The results are shown in Table 3, where the dose per emitted photon is computed from the experimental data and compared to the computer calculated value. In both cases, the measurements and calculations agree, within the uncertainties.

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Table 2. Results of phantom scans

Source-Phantom Description	True Activity (mCi)	Calculated $\bar{A}$ (mCi)	Calculated A (mCi)	Error (%)
Plane Source				
Homogeneous	1.35	1.37	1.37	+1.4
Cube Source				
Homogeneous	1.28	1.41	1.29	+0.9
Cube Source	1.40	1.54	1.41	+0.8
Heterogeneous	1.28	1.41	1.29	+0.9
Irregular Source	0.689	0.762	0.709	+2.9
Heterogeneous	0.664	0.729	0.679	+2.3

Table 3. MR. ADAM results

Source	Dose ($X10^{-14}$ rads/photon)		
Organ	From Measurements	Computer-Calculated	Meas/Cal
Liver	5.37 ± 6.5%	5.305 ± 12.2%	1.01
Spleen	3.66 ± 9.4%	4.033 ± 26.1%	0.91

PATIENT RESULTS

The patient results are shown graphically in Figure 3; the dashed lines indicate a 30% difference in either direction. Each measured dose was found from TLD readings and felt to have an overall uncertainty of ±5%, while each calculated dose was found from the cumulated activities and S-factors, and felt to have an overall accuracy of ±10%. In most cases, the discrepancy is ±30% or less, indicating that the model does apply reasonably well to patients. Further, in all cases where the calculated dose was appreciably larger than the measured dose (up to 67% larger), the patient was either larger than the model (taller and heavier) or else the active liver was abnormally small due to disease (cirrhosis). In all cases where the calculated dose was smaller (down to only 45% of measured dose), the patient was smaller than the model. This is as expected, since any discrepancy would depend upon the amount of attenuating material between the liver and TLD's, relative to that of the model.

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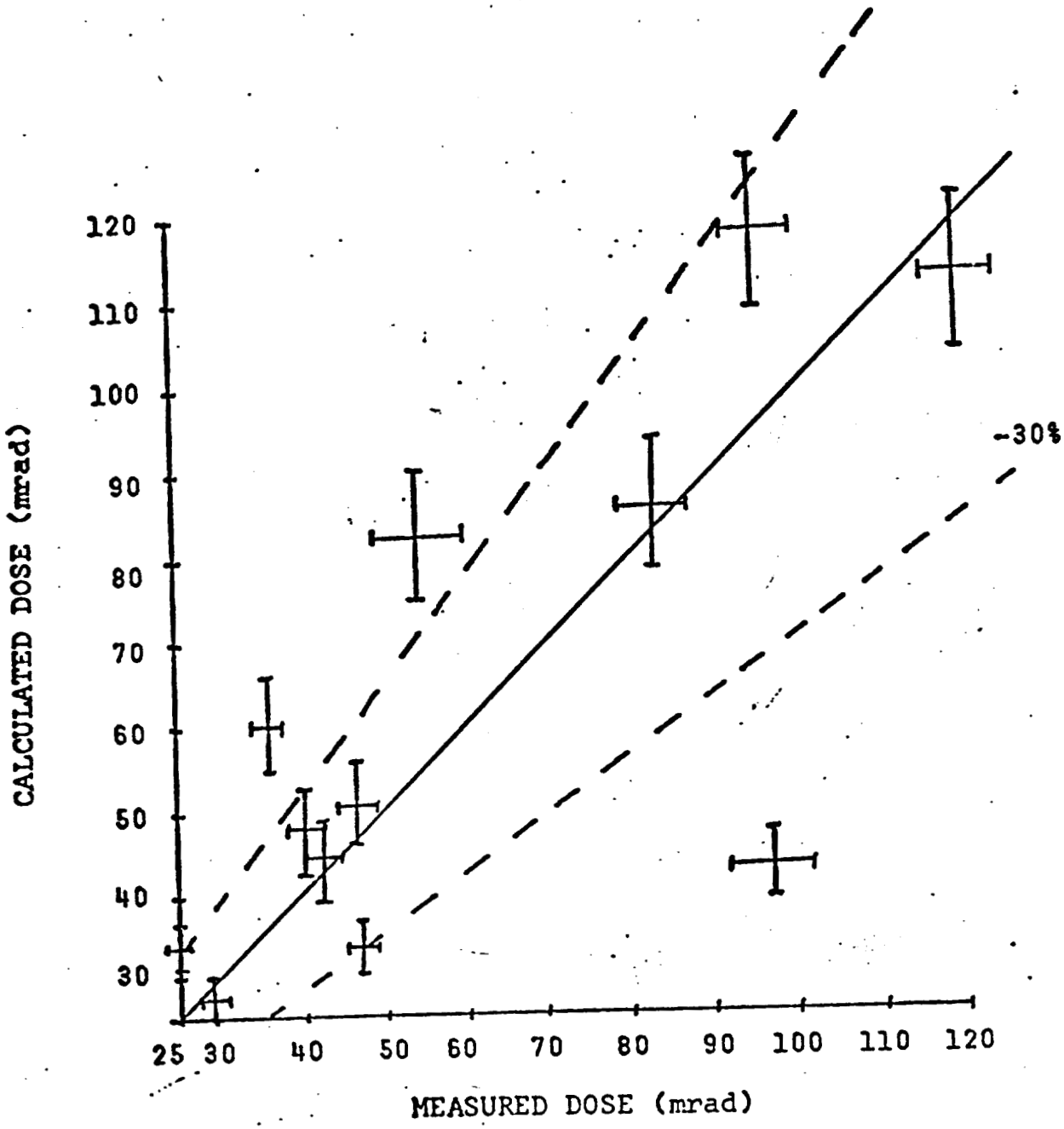

Figure 3. Patient results.

REMARKS

While this study gives some idea of how well the model S-factors apply to real patients, it is only the beginning of a needed evaluation, since comparable results might not exist for other source-target configurations. But some general conclusions can be drawn. For example, with higher photon energies, geometric differences between the model and patients will have less effect, because the attenuation is less. Also, for much lower photon energies, geometric differences will again be less important, because most of the photon energy will be absorbed by the source organ(s). But for intermediate energies, geometric differences will be more important, especially for those energies somewhat less than 140 keV. The should be investigated, because some commonly used isotopes (^{131}I, ^{133}Xe, ^{197}Hg) emit photons of energy 70-80 keV.

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The general approach used here, of defining an external target region suitably positioned close to a major source organ, could be used to evaluate other source distributions. However, one must be sure that the target dose is large enough to be accurately measured (≥20 mrad for LiF TLD's), and that the cumulated activity can also be accurately determined. The method of activity determination used here may be useful, especially if there are multiple source organs to consider, because it allows one to calculate the activity of resolved sources separately. This method is shown in more detail in Table 4, because it involves approximations which become less valid as the amount of photon attenuation increases.

Table 4. The method of activity determination

$$\bar{A}/A_T = [\sum_j^m E_j \exp(-\mu dj + \mu tj/2 + \mu z_{j1})]^{\frac{1}{2}} [\sum_j^m E_j \exp(-\mu t_j/2 - \mu z_{j1})]^{\frac{1}{2}} \exp(\mu d/2)$$

where $E_j = t_j \sinh(\mu tj/2)/(\sum_k^m tk)\ (\mu t_j/2)$

and A_T = true source activity.

Schwarz Inequality is

$$\sum_j^m X_j^2 \sum_j^m Y_j^2 = \sum_j^m X_j Y_j, \text{ if } (Y_j/X_j)^2 \text{ is independent of } j$$

$$\sum_j^m X_j^2 \sum_j^m Y_j^2 > \sum_j^m X_j Y_j, \text{ otherwise.}$$

If one defines $A = \bar{A}/\sum_j^m E_j \exp[(-\mu dj + \mu d)/2]$,

then

$$A = A_T, \text{ if } \exp(-\mu dj + \mu tj + 2\mu z_{j1}) \text{ is independent of } j \quad (1)$$

$$A > A_T, \text{ otherwise} \quad (2)$$

Assuming that the detectors are well-collimated, the source is uniform, and the attenuation of photons by the scattering medium can be described by a simple exponential function leads to the first equation shown, where $\bar{A}$, A, d, d_j, and t_j are the same as in Table 1, and the z_{j1} are the set of coordinates of the bottom of the source organ. Because this equation could not be evaluated directly, the Schwarz Inequality was applied, leading to equation (1) and inequality (2). The condition for (1) to be valid is a geometric one, since it relates the source and total body thicknesses. By analysis of ultrasonic liver scans, we found that d_j, t_j, and z_{j1} were such that this condition was not met. But if μ is small enough, the exponential will still not vary by much, regardless of how the d_j, etc. vary. This is important because it can be proven that the Schwarz Inequality is still an equality for first order violations of that condition; the inequality is due to second order effects. Again using the ultrasonic scans to estimate a set of d_j, t_j, and z_{j1} values, we found μ to be small enough in this study that (1) was only 5% in error. Further, this error is systematic, so it could be accurately corrected for. For higher energy photons (smaller μ) it follows that this error would be less, while for lower energy ones, the error would slowly increase with increasing μ until second order effects became pronounced. Then, the error would increase rapidly with increasing μ, and this method would fail.

To quantitatively estimate the applicability of this method for lower photon energies, the same set of d_j, t_j, and z_{jl} were used as before, along with values of μ corresponding to 80 keV, and then 30 keV photons. For 80 keV, the error of (1) was about 15%, and for 30 keV, it exceeded 50%. Thus at 80 keV, this method should be adequate, because a 15% correction is not too large to make accurately. But for 30 keV, this is not true, and this method will probably fail.

Because of the tedium of evaluating these equations with sets of d_j, t_j, and z_{jl}, it is desirable to use a simpler equation to check this method. This can be done by simply (with a computer) computing the percent standard deviation from the mean of the set of values $\exp(-\mu d_j/2)$, $j=1, \ldots m$. If this percent standard deviation is appreciable ($\geq 10\%$), then μ is likely not small enough to allow (1) to be valid. But if it is small (observed to be 4% in this study), then μ is certainly small enough to mask the variation of the d_j values, and is likely to be small enough to assure that (1) is approximately correct (provided that the numbers $t_j - 2z_{jl}$ do not fluctuate wildly, which is unlikely since they are proportional to the set of mid-depths of the source organ). Also, if this percent standard deviation is small, one can approximate $\exp[-\mu(d_j - d)/2]$ by unity, which results in the equation for A given in Table 1.

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