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IN VIVO GAMMA LUNG MEASUREMENTS -
A MATHEMATICAL MODEL

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INTRODUCTION

The in vivo detection and measurement of radioactive deposits in body organs is afforded by the external scintillation spectroscopy of emitted gamma radiation. A low-background facility for measuring lung burdens of uranium, thorium, and other nuclides is in operation in the Y-12 plant.⁽¹⁾ In brief, this in vivo unit consists of: (1) a low-background counting chamber constructed of steel floors, walls, and ceiling; (2) a nine-inch diameter NaI (TI) crystal detector mounted on a movable carriage assembly, and (3) a 256-channel pulse height analyzer. To measure the gamma radiation emitted by radioactive lung deposits, an examinee, thoroughly washed and wearing radioactively clean clothing, is positioned on a cot or in a chair with the lowered detector centered over the chest cavity. During a prescribed counting period (usually 20 to 30 minutes) the measured gamma activity is electronically recorded according to energy level. Estimates of background radiation and the normal in vivo spectrum, part of which is due to the presence of such nuclides as Cs-137 and K-40 in the examinee's body, are subtracted from the gross measured spectra. Then, the observed count rate is converted by means of an empirical calibration factor into an estimate of the mass of material in the body.

Two major problems must be overcome for effective use of in vivo gamma counting. First, in order to define the net radiation spectrum which can be attributed to the presence of uranium^(a) in the lungs, the total background spectrum and the examinee's normal in vivo spectrum must be known. Contribution to the background spectrum made by the chamber and by radioactive material adhering to the examinee's skin can be minimized by careful construction of the counting chamber and rigorous preparation of the examinee prior to counting. The normal in vivo spectrum, however, varies from person to person because of differing body geometries and variations in the body content of nuclides the concentrations of which are a function of weight, diet, and exposure to fallout from atmospheric contamination. A system by which the normal in vivo spectrum for any individual can be predicted has been developed by Cofield.

The second major problem is relating the net radiation spectrum to the actual lung burden of radioactive materials, in particular U-235. Cofield devised a general correlation between an observed count rate and the mass of uranium in the lungs using a Masonite phantom of the chest. Small uranium sources were placed at various positions within an imaginary 6-inch disc centered in the midplane of the phantom. The disc, in turn, was located about 4 1/2 inches from the detector face. Since the count rate of a source varied with its position on this circle, a weighted average ratio of count rate to uranium mass (i.e., an instrument calibration) was thus determined. Correction factors for the calibration factor as a function of chest thickness were also calculated.⁽¹⁾

(a) The problems exist for all radioactive nuclides; uranium is cited in this paper because of its importance in Y-12 operations.

In theory, the mass of uranium contained in the lungs, M_U , is directly proportional to the intensity of the observed radiation, I_m , at the energy level of interest. Thus,

$$M_U = \left(\frac{1}{S \cdot F} \right) I_m \quad (1)$$

where "S" is the specific activity of one or more nuclides and "F" is that fraction of the radiation which is measured by the detector (i.e., the measurement efficiency). In this analysis, the detector is considered to count all incident gamma photons. Radiation emanating from deposits in the lungs may be absorbed or scattered in several intermediate body tissues. Only that portion of the gamma radiation which leaves the body and is geometrically intercepted by the detector is measured. Thus, "F" for a point source is given by:

$$F = \frac{I}{I_0} = \frac{A e^{-\sum_i \sigma_i l_i}}{4\pi d^2} \quad (2)$$

where

I is the radiation intensity at the detector,

I_0 is the radiation intensity of the source,

σ_i is the attenuation coefficient for the i th tissue,

l_i is the thickness of the i th tissue,

A is the area of the detector normal to the flux, and

d is the distance between the source and the detector.

It is clear that "F", given by Equation 2, and hence the calibration factor (the correlation between net count rate and lung burden - $1/S \cdot F$ in Equation 1) is a function of the geometry of the system (including the examinee and the detector) and the distribution of uranium in the lungs, because these factors influence both the " d " and " $\sum_i \sigma_i l_i$ " terms in Equation 2.

In order to examine the effect of variation in lung geometry and distribution of radioactivity on the counting efficiency, F , and also to develop a better understanding of in vivo measurement techniques, a mathematical description of the in vivo counting process was devised. The system under consideration was defined as the detector face and all of the chest cavity between the top and bottom of the lungs. This region of the body was positioned in a three-coordinate space, as shown in Figure 1. The chest was partitioned into one-inch cubes by taking one-inch increments along each major axis.

Figure 1 THE COORDINATE GRID SUPERIMPOSED UPON CHEST

For the simulation analysis, hypothetical chest cavities were created for three very different persons — one relatively small, one about average, and one quite large. Chest X rays and caliper measurements of actual people fitting these sizes were employed as guides to the creation of reasonable data.^(a) With the X-ray pictures, the "X-Z" planar dimensions and positions of the lungs, heart, and ribs were determined. External caliper measurements provided an estimate of chest depth (along the Y coordinate) of the chest. References on human anatomy (2, 3) were employed for developing the dimensions of the bones and major organs. Cross sections at one-inch intervals along the "Z" axis were postulated using photographs of horizontal cross sections contained in a report by Eycleshymer and Shoemaker (2) and scaled to the external dimensions. Therefore,

the composition of tissues in each one-inch square (in the "X-Y" plane) was determined at intervals of one inch along the "Z" axis.

Four types of body tissue—lung, bone, heart, and soft tissue (muscle, cartilage, and fat)—were considered as possible constituents of each cube. The "X-Y" area fraction of each component tissue associated with a grid point was assumed to represent the volume fraction of the cube, of which the grid point was the center. Each cube that contained greater than 50 percent lung tissue was postulated as containing a point source of monoenergetic radiation.

The fraction of gamma activity emitted by each source point in the system and intercepted by the crystal detector can be calculated by Equation 2. In the computations, the distance between source and detector, and the location and type of intermediate tissues must be known. Thus, the path of radiation from every source point to each of many detector points had to be ascertained for in vivo calculations. Consequently, the model of in vivo radiation measurements was programmed for solution on a high-speed digital computer using the data on position and composition of each cube in the system. These data were encoded

(a) Note that no attempt was made to simulate an actual person.

and put onto magnetic tape for use with the computer program. The computations were made for two distinct sets of tissue attenuation coefficients. In the first case, absorption coefficients were used, and in the second, total linear (absorption plus scattering) coefficients were employed. The values of these coefficients, which were taken from Spiers' data⁽⁴⁾, are presented in Table D-1, Appendix D. The program is considered in more detail in Appendix C, and a sample calculation is presented in Appendix D.

The versatility of the computer simulation for a qualitative study of *in vivo* gamma radiation measurements should be emphasized. The relative effect of each variable on the counting efficiency, or calibration factor, for a given detector location can be investigated individually. Wider ranges of each variable can be studied than would be possible in a purely experimental approach in which the source distributions available for experimental study are limited. Furthermore, numerous detector positions and the use of combinations of detectors can be easily investigated in order to ascertain the method of counting in which the counting efficiency, F , is least affected by variations in body geometry and source distribution.

Counting efficiencies were determined for a hypothetical person slightly greater than average in size and weight with a nine-inch diameter detector positioned at the right and left sides, and front and back of the chest. In addition, flux distributions on the detector face were computed for each of these geometries. The effect of body size on counting efficiency with the detector at the front of the chest was determined using geometries for two additional individuals, one extremely large and one very small. By an examination of the results, insight has been obtained to show the most effective means for detecting and estimating lung burdens of radioactive material.

SUMMARY

In vivo gamma radiation measurements are becoming increasingly important for the detection and estimation of radioactive deposits in the human body. One problem confronting this method of radiation measurement is relating the net observed radiation spectrum to the burden of radioactivity in the body or its organs (i e, calibrating the instrument). In order to gain some insight into this problem, a mathematical, computer-oriented simulation of the in vivo measuring process was devised. Chest cavities for persons of three sizes — small, average, and very large — were synthesized in a three-coordinate space comprised of one-inch cubes and including a nine-inch-diameter crystal detector. Data, describing the tissue composition of each cube in the body and the characteristic radiation attenuation for each tissue-type, were coded for use with a program on a high-speed digital computer. Subsequently, efficiencies for measuring radiation emitted by numerous "point sources" of enriched uranium were calculated.

Measurement of the 186-Kev photons from U-235 with the detector positioned at the back of an average person gives a maximum theoretical efficiency of 12 percent, but an actual efficiency of between 4 and 9 percent. Efficiencies for making measurements at the front and sides are 20 to 30 percent smaller than these values. In obtaining these results, uniform distribution of the radioactivity in the lungs was assumed. Major attenuating organs are the heart for the anterior and the spine for the posterior detector positions. The data show that measurement efficiencies are higher for small persons (whose lungs are, in general, closer to the detector) than for larger ones. From the results it is also apparent that the instrument calibration (measurement efficiency) should be least sensitive to individual size if the detector is positioned at the back of the chest. This is primarily due to the fact that more than 60 percent of the lung tissue is in the posterior half of the chest cavity. When the detector is positioned at either side of the chest, the counted radiation emanates primarily from the nearest lung. If the radioactivity is concentrated in certain regions of the lungs, rather than being uniformly distributed, measurement efficiencies are very sensitive to the location of the sources with respect to the detector and, thus, to the size of the examinee.

The computer simulation used in this study has been shown to be a fairly accurate representation of in vivo gamma counting. This technique has application in the development of a better understanding of in vivo measurements although it cannot, at present, be employed for calibrating in vivo counting instruments.

EXPERIMENTAL RESULTS

INTRODUCTION

Data on in vivo measurement efficiency has been obtained for two general situations: (1) uniform distribution of radioactive material throughout the lungs, and (2) nonuniform deposits. The effects of individual size and geometry, and of detector position on the measurement efficiency were determined for these two categories. Radiation flux distributions on the detector face were computed in some cases.

COUNTING EFFICIENCIES WITH UNIFORMLY DISTRIBUTED RADIOACTIVITY IN LUNGS

Parameter Study for Nearly Average Individual

Counting efficiencies were calculated for a hypothetical person, slightly above average in size and weight, with the nine-inch-diameter detector positioned at the front, back, and sides of the chest. Assuming that 186-Kev monoenergetic radiation emanates from the deposit, measurement efficiencies were calculated as listed in Table I for the following circumstances:

1. The body does not absorb or scatter any radiation and is therefore transparent to it — the coefficients are equal to zero.
2. Only radiation which is neither absorbed nor scattered is counted — linear absorption (absorption plus scattering) coefficients are employed.
3. All unabsorbed radiation (including degraded photons reaching the detector) is measured — absorption coefficients are used.

The significance of the efficiencies determined for these conditions is: (1) when the body is assumed not to attenuate any of the radiation, a maximum efficiency, based upon the system geometry, is obtained; (2) if only radiation reaching the detector at 186 Kev were counted, the lowest possible efficiency would be computed, and (3) if both scattered and unabsorbed radiation (with energy between 0 and 186 Kev) is counted, the highest realistic measurement efficiency is obtained. Assuming that the U-235 emits gamma radiation monoenergetically at 186 Kev, measurements at the Y-12 Plant facility would be made in the lower half of the 165 to 215 Kev band width. Therefore, a reasonably realistic measurement efficiency would lie between those obtained for Cases 2 and 3 previously listed. Although uranium emits gamma photons with a distribution of energies having a mean of 186 Kev, the calculations were greatly simplified by assuming a monoenergetic source. However, the results obtained herein should approximate the more realistic situation.

Table I
IN VIVO MEASUREMENT EFFICIENCIES AS A FUNCTION OF DETECTOR LOCATION
(Fractional Efficiency)

Detector Position	Attenuation Coefficient				
	Absorption plus Scattering			Absorption	
	Zero	Individual ⁽¹⁾	Average ⁽²⁾	Individual ⁽¹⁾	Average ⁽²⁾
Front	0.092	0.028	0.028	0.069	0.071
Right Side	0.101	0.037	0.036	0.079	0.080
Back	0.120	0.042	0.044	0.093	0.096
Left Side	0.086	0.024	0.021	0.063	0.062

⁽¹⁾ Individual coefficients used for each tissue type.

⁽²⁾ A weighted average coefficient employed for all tissue types.

Referring to Table I again, the geometrically maximum measurement efficiency is about 12 percent where the detector is in the back position. This value is 20 to 30 percent greater than the efficiencies attainable with the detector at the other locations. A realistic maximum efficiency lies between 4 and 9 percent. Included in Table I are efficiencies computed for "average" attenuation coefficients. These coefficients are weighted average values for the entire chest as determined from the composition of this region of the body, and the one average volume-weighted coefficient is employed for each of the four tissue types.

Figure 2. COUNTING EFFICIENCY AS A FUNCTION OF PLANAR DEPTH, Y, FOR FRONT AND BACK DETECTOR POSITIONS.

The counting efficiency was computed for one-inch-thick planes perpendicular to the Y axis (i.e., the dimension of chest depth) and extending across the chest. Plots of the efficiencies for both the front and back detector positions are shown in Figure 2 as a function of planar location. In addition to illustrating a rapid decrease in flux with increasing distance between source and detector (the inverse distance-squared effect), the data show that generally higher fluxes are obtainable with the detector positioned at the back. The reason for this is the proportionately higher mass of lung tissue at the back of the chest cavity. Figure 3 shows that, for the hypothetical individual used in the calculations, about two-thirds of the lung tissue was located in the posterior half of the

Figure 3. LUNG TISSUE VOLUME AS A FUNCTION OF PLANAR DEPTH, Y.

chest. Lung geometry which is governed by the location of the heart, the gradual taper of the chest at the shoulders, and the forward and upward slope of the liver immediately below the lungs, largely accounts for this condition (see Figure 1). Therefore, positioning the detector at the back of the chest utilizes to advantage the geometry of the human anatomy.

The chest was similarly divided into one-inch-thick planes in the Y-Z plane (referring to Figure 1). The counting efficiencies for the lung tissue in each of these planes were computed and are plotted in Figure 4. The higher flux contribution from the right lung results from a combination of the larger tissue volume of this organ, as shown in Figure 5, and shielding by the heart on the left side. A higher counting efficiency for the back detector position is again indicated from Figure 4.

Figure 4. COUNTING EFFICIENCY AS A FUNCTION OF PLANAR POSITION, X, FOR FRONT AND BACK DETECTOR POSITIONS.

Figure 5. LUNG TISSUE VOLUME AS A FUNCTION POSITION ACROSS BODY, X.

The counting efficiencies with the detector positioned at either side are less than at the back, and the right side yields higher count rates than the left side (see Table I). It is evident from Figure 6 and Table II that the major fraction of the count rate measured at the

Figure 6. COUNTING EFFICIENCY AS A FUNCTION OF PLANAR POSITION, X, FOR SIDE DETECTOR POSITIONS.

Table II
IN VIVO MEASUREMENT EFFICIENCIES ON THE RIGHT AND LEFT SIDE POSITIONS

Detector Position	Right Lung Contribution		Left Lung Contribution	
	Zero Coefficients (%)	Absorption Coefficients (%)	Zero Coefficients (%)	Absorption Coefficients (%)
Right Side	83	94	17	6
Left Side	27	9	73	91

side positions is obtained from the adjacent lung; the inverse law effect plus some tissue attenuation reduces the gamma flux from the opposite lung to a comparatively small value. The counting efficiency for the detector on the left side is understandably lower than that on the right side because only about 40 percent of the lung tissue is contained in the left lung (see Figure 5). The flux contributions from X-Z planes, one-inch thick, through the depth of the chest are plotted in Figure 7, and show only slightly higher counting efficiencies from the tissue in the posterior section of the chest.

Contours of radiation flux over the detector face for the four positions under consideration are presented in Figures 8 and 9. In the frontal position, a high-intensity region appears on the detector directly opposite the right lung; lack of a peak region on the left side is indicative of both the absence of radioactive lung tissue and higher attenuation in the heart region. In contrast,

the detector face position at the back produces a high-intensity region from each lung. In this case, the somewhat lower count rate in the middle of the detector may be due to either the central position of the highly absorptive spinal column, or the absence of lung tissue in the center of the chest, or both. The flux patterns for measurements taken at the side positions contain a single high-intensity region slightly below the center of the crystal (clearly a function of detector position), and nearly symmetrical flux contours.

Figure 7. COUNTING EFFICIENCY AS A FUNCTION OF PLANAR POSITION, Y, FOR SIDE DETECTOR POSITIONS.

Since a nine-inch-diameter crystal might

Figure 8. CONTOURS OF GAMMA RADIATION FLUX ON DETECTOR FACE FOR FRONT AND LEFT SIDE POSITION.

Figure 9. CONTOURS OF GAMMA RADIATION FLUX ON DETECTOR FACE FOR BACK AND RIGHT SIDE POSITION.

not be conveniently positioned at either the left or right sides of the chest, counting efficiencies were determined as a function of crystal diameter, as shown in Table III. By reducing the diameter to five inches, the counting efficiency is reduced to slightly less than half. However, the positioning of two smaller crystals on either side of the body and counting simultaneously would produce about the same efficiency as with a single large crystal.

Table III
COUNT RATE AS A FUNCTION OF DETECTOR DIAMETER FOR SIDE POSITIONS

Diameter (in)	Area (in ²)	Right Side		Left Side	
		Zero Coefficients	Absorption Coefficients	Zero Coefficients	Absorption Coefficients
9	61.6	0.101	0.037	0.086	0.024
8	50.3	0.087	0.030	0.073	0.022
7	38.5	0.071	0.025	0.059	0.018
6	28.3	0.058	0.021	0.049	0.016
5	19.6	0.045	0.016	0.037	0.013

Effect of Body Size on Counting Efficiency

Measurement efficiencies have been determined for three sizes of adults — one extremely small, one about average, and one very large. Pertinent statistics on these hypothetical persons and the computed efficiencies for the front detector position are presented in Table IV.

Table IV
IN VIVO GAMMA COUNTING EFFICIENCIES FOR THREE BODY SIZES FOR DETECTOR IN FRONT OF CHEST

Individual	Approximate Weight (lbs)	Max Chest Thickness (in)	Max Chest Width (in)	Lung ⁽¹⁾ Volume (in ³)	Counting Efficiency		
					$\alpha^{(2)}$	$\beta^{(3)}$	$\gamma^{(4)}$
A	90	7	12	242	0.137	0.037	0.078
B	180	9	12½	242	0.092	0.028	0.069
C	235	12	14	296	0.092	0.020	0.050

⁽¹⁾ Determined by counting number of one inch cubes containing lung tissue.

⁽²⁾ α is the case in which zero attenuation in tissues is assumed.

⁽³⁾ β is the case in which absorption and scattering coefficients were employed in the calculations.

⁽⁴⁾ γ is the case in which only absorption coefficients were employed.

The results show that there is probably no unique efficiency for a given chest geometry (as evidenced by Case α in the table). However, there is a general trend, under actual in vivo conditions (Cases β and γ), of increasing counting efficiency with decreasing body

size. This trend is a consequence of the nearness of the bulk lung tissue (and hence the radioactive deposit) to the detector for the smaller person. The analysis is not straightforward, however, because both the lung volume and the counting system geometry are functions of body size. This report does not include a detailed study of the interrelationship of system geometry and chest dimension on counting efficiency.

From the results of the previous section it is inferred that the in vivo counting efficiency should be less sensitive to variations in body and system dimensions if the detector is placed at the back of the chest.

COUNTING EFFICIENCIES WITH NONUNIFORM RADIOACTIVE LUNG DEPOSIT

Under particular conditions of short-term exposure, inhaled radioactive particulates are likely to deposit nonuniformly in the lungs. (5, 6) In routine in vivo measurements, however, a standard distribution of uranium in the lungs is assumed and a constant, empirical counting efficiency is employed since the determination of the actual distribution and geometry in each case is virtually impossible. To demonstrate the magnitude of errors associated with these assumptions, counting efficiencies were computed for limited-size sources concentrated in four regions of the lungs. The four source volumes and locations are: (1) a two-cubic-inch parallelepiped located at the top of either lung; (2) a four-inch-diameter disc, one-inch thick, whose centroid is located at the center of gravity of either lung; (3) six-cubic-inch parallelepipeds positioned at the central, posterior region of the lungs, and (4) a six-cubic-inch parallelepiped at the central, anterior region of the right lung. Data on counting efficiencies were obtained for the four source locations cited using small, average, and large-sized individuals. The results are presented in Table V which also presents the counting efficiencies computed for a uniform source distribution.

Geometry Effects on Counting Efficiency

The effects of in vivo measurement geometry on counting efficiency are obtained from data under the column "Zero Coefficient", in Table V, for which circumstance the body is assumed to be transparent to the radiation. The uniform distribution yields efficiencies of almost 14 percent for the small individual (A) and about nine percent for the average and large persons (B and C). Four-inch discs located in the center of each lung, which could be used for calibrating the in vivo instrument with a manikin, have approximately the same efficiencies. Slightly higher values were obtained for the source in the right lung than in the left. Unless inaccuracies in locating the sources and detector in the computer program can account for this trend in efficiencies, the differences cannot be explained.

Efficiencies slightly lower than those for the discs and a uniform distribution occur with sources positioned at the top and back of the lungs. However, the lowest efficiency for the small individual resulted from the source at the top of the lungs, whereas the back position

Table V
IN VIVO COUNTING EFFICIENCIES FOR SEVERAL DISTRIBUTIONS OF LUNG DEPOSITS
(Front Detector Position)

Case	Deposit Location and Size	Attenuation Coefficients				
		Zero			Absorption Plus Scattering	Absorption
		A ⁽¹⁾	B ⁽²⁾	C ⁽³⁾	B	B
1	Top of Lungs (1" × 1" × 2" Parallelepiped)					
	Left Lung	0.059	0.076	0.067 ⁽⁴⁾	0.016	0.053
	Right Lung	0.063	0.094	0.076	0.023	0.068
2	Center of Either Lung (4" D Disc, 1" Thick)					
	Left Lung	0.111	0.081	0.084	0.019	0.058
	Right Lung	0.168	0.108	0.101	0.037	0.085
3	Back of Either Lung (1" × 2" × 3" Parallelepiped)					
	Left Lung	0.109	0.052	0.049	0.007	0.033
	Right Lung	0.114	0.059	0.050	0.012	0.041
4	Front of Either Lung (1" × 2" × 3" Parallelepiped)					
	Left Lung	--	--	--	--	--
	Right Lung	0.366	0.201	0.261	0.089	0.167
5	Uniform Distribution	0.137	0.092	0.092	0.028	0.069

⁽¹⁾A is the small, 90 pound individual.

⁽²⁾B is the average, 180 pound individual.

⁽³⁾C is the large, 235 pound individual.

⁽⁴⁾Computed for 1" × 1" × 1" deposit.

proved to yield the lowest values for the larger persons. Sources located near the front of the lungs have the largest effect on counting efficiency, as shown for Case 4 in Table V. Radioactivity in the lungs nearest to the detector is counted with efficiencies two to three times those of the uniform distribution.

Attenuation Effects on Counting Efficiency

Counting efficiencies computed using tissue attenuation of the gamma photons are more realistic than the data compiled for the theoretically radiation-transparent body. Results computed for the average individual are presented in Table V for the two extreme cases: (1) only unabsorbed and unscattered radiation, at 186 Kev, is counted (absorption plus scattering coefficients are used), and (2) all unabsorbed and scattered radiation is counted (absorption coefficients are employed). The former situation tends to give a lower-than-expected value while the latter coefficients give higher counting efficiencies. For an assumed uniformly distributed deposit of radioactivity in the lungs, the counting efficiency ranges from two to four percent for the largest individual to 3.7 and 7.8 percent for the

smallest. The efficiencies calculated for the four-inch-diameter discs varied from the uniform distribution calculations by a factor of 1.3. Less agreement is thus shown than might be expected from geometrical considerations alone. The efficiencies computed for the disc-type source in the left lung are slightly lower than those computed for the right-lung location. The fact that Individual B differs by a factor of two under conditions of attenuation indicates the attenuation effect of the heart for the left-lung position as compared to the relatively small lung-tissue attenuation on the right side.

Considerable variability in counting efficiency is demonstrated as the source location is shifted throughout the lung region. Of particular importance are the significantly high efficiencies attained with the radioactive source positioned at the front of the lungs. Since the values shown in the table approach those calculated for the "no-attenuation" case, it is shown that small attenuation occurs in the front wall of the body and that distance is a primary parameter governing the counting efficiency of a specific source position.

DISCUSSION

INTRODUCTION

The validity of the results obtained from the in vivo gamma measurement simulation is determined by: (1) the accuracy of the computations, and (2) the soundness of the model employed. Since the calculations (as outlined in Appendix D) involve numerous approximations and assumptions concerning such items as grid size and tissue attenuation coefficients, various types of uncertainties and inaccuracies are present in the calculations. Some of the most important problems are discussed in the sections that follow.

ACCURACY OF THE SIMULATION

The accuracy of the computations was checked by simulating a simple radiation experiment (see Appendix A). In the laboratory, a point source was positioned about one foot from a nine-inch-diameter sodium iodide detector, and the count rate was determined for varying thicknesses of Masonite shielding intermediate between the source and detector. The experiments were simulated using the computer program from which the ratio of intensity with "t" shielding thickness to that with no shielding, $I_t/I_0 = 0$, was determined. An analytical solution to the attenuation equations was found for the relatively simple geometry involved. The ratios, $I_t/I_0 = 0$, determined by the three methods showed excellent agreement ($\pm 5\%$) for thicknesses up to nine inches of Masonite. The significance of this analysis is that added confidence may be placed in the computer program for homogeneous shielding and a simple source. However, for the more complicated human anatomy with varying tissue composition and multiple point sources, the relatively small errors demonstrated with a point source may become compounded.

THE EFFECT OF GRID SIZE ON THE ACCURACY OF THE SIMULATION

In the limit, as the grid size is reduced to infinitesimally small increments, the simulation should be exact providing that the tissue attenuation coefficients are accurate. Because computer calculation time, and hence the cost, increases dramatically with a decrease in grid size^(a), it is practical to use as large a grid as possible while maintaining the desired accuracy. In all calculations performed for this study, a one-inch, three-dimensional grid was employed. As a test of the reliability of the results obtained under these conditions, part of the hypothetical, average individual was sectioned into cubes one-half inch on a side. In the smaller grid there were 2^3 or 8 times as many source points in a given volume as in the one-inch grid. The one-half-inch grid had an added advantage of representing

(a) For example, if the grid size is halved, the computation time is increased 2^5 or 32 times: 2^3 for the number of source points and 2^2 for the number of detector points.

sources better at the outer surfaces of the lungs. Counting efficiencies were determined for the test grid size using the computer program (see Appendix B).

Counting efficiencies, calculated for about half of the person's lung volume, were about 15 percent higher with the one-inch grid than with the smaller coordinate system. At the boundary of the lung tissue discrepancies were quite large. Thus, the results of the computations based on the one-half-inch grid indicate significant bias in the reported counting efficiencies. Although the data from the one-inch coordinates are slightly low, the results presented in this paper appear to be adequate for providing insight into in vivo gamma counting.

SOURCE ENERGY AND TISSUE ATTENUATION COEFFICIENTS

Uranium-235 emits gamma radiation with a spectrum of energies. Two prominent emission energies are at 90 and 186 Kev, which are mean values for a distribution of photon energies at these levels. The higher energies are most valuable for in vivo monitoring for enriched uranium and, at Y-12, a range of 165 to 215 Kev is used for measuring the higher-energy gamma radiation. To simplify the computations in this study, a monoenergetic source at 186 Kev has been assumed. This approximation appears to be reasonable because the absorption and scattering coefficients of the body tissues change only slightly over the 165 to 215 Kev range of energies.

The absorption and scattering coefficients for bone, fat, and muscle were taken from data by Spiers⁽⁴⁾. The linear absorption (absorption plus scattering) coefficients were taken directly from the data, but the coefficient for lung tissue had to be estimated. This was accomplished by taking the density of the tissue in the lung, and the density of air (from the void volume), and computing a weighted average linear absorption coefficient. The resultant absorption coefficient was assumed representative of lung tissue from the values for air and muscle. Although there may be some inaccuracy to this method of estimation, the relatively low amount of attenuation in the lungs did not warrant a more precise determination. Heart tissue was assigned the same linear absorption coefficient as muscle.

The absorption-only coefficients were determined indirectly from energy absorption coefficients, σ_{ea} , which may be represented by

$$\sigma_{ea} = \sigma_a + f\sigma_s$$

where " σ_a " and " σ_s " are the absorption and scattering coefficients, and "f" is the fraction of gamma ray energy converted to kinetic energy of recoil electrons in a single collision. By taking Spiers data, simple computations showed that $f\sigma_s$ was small with respect to σ_a at 186 Kev and, hence, the energy absorption coefficients given in the reference could be employed as absorption coefficients. Values for the lung tissue were computed in the same

manner as for the total linear absorption coefficients.

The use of absorption-only coefficients for computing counting efficiencies presumed that all nonabsorbed gamma photons (including those scattered and degraded to lower energies) which were incident upon the detector were counted. Thus, the spectrum of counted gamma radiation would range between 0 and 186 Kev. Since the Y-12 procedure is to count between 165 and 215 Kev, the computed efficiencies using linear absorption coefficients would be higher than might be expected in an actual facility. Counting efficiencies determined with linear absorption coefficients are based on counting nonabsorbed and non-scattered gamma photons. Thus, radiation which is measured by the detector must have the same energy as the source — in this case 186 Kev. Since some scattered radiation is counted in actual practice, efficiencies computed using these coefficients are necessarily lower than those attained in actual in vivo measurements. Whereas, the exact simulation of in vivo counting is difficult and costly, the employment of linear absorption and absorption-only coefficients determine the upper and lower limits of actual counting efficiencies.

DISTRIBUTION OF RADIOACTIVE DEPOSITS IN THE LUNGS

For routine in vivo measurements, uniform distribution of radioactivity in the lung tissue is assumed. In reality, nonuniform deposits should usually result from inhalation exposures. Fish⁽⁷⁾ found that after exposing dogs to UO₂ aerosols, nonreproducible, nonuniform deposits in the lung tissue resulted. Morrow⁽⁶⁾ in a theoretical study has demonstrated that both physical and physiological factors govern the deposition of inhaled aerosols in the lungs.

Particle size and aerosol concentration as well as breathing rate and tidal volume are important variables in dust deposition. The composition of respiratory tract air changes continuously, and, as Morrow wrote, "this implies that in different parts of the lungs, the particles of dust inhaled will have different residence times and that deposition will not be uniform". The distribution of radioactive particulates in the lungs is also a function of the history of exposure. For a single inhalation, a nonuniformly distributed deposit would be expected. In the case of chronic exposure, the variation in conditions, both physical and physiological, from inhalation to inhalation would probably result in a fairly uniform distribution of radioactivity in the lung tissue. In addition, the processes for the elimination of materials from the lungs would achieve essentially steady-state conditions. Casarett⁽⁵⁾ has shown that a result of chronic exposure to some nonsoluble radioactive materials results in an accumulation in the lymph nodes near the lungs. Concentrated sources of radioactivity in these regions would effect the counting efficiency of an in vivo instrument.

In selecting several locations for limited-size radiation sources, no attempt was made to simulate any particular geometry or situation. Rather, the choices were made to represent extremes in radiation counting geometries and provide an estimate of the variation in

counting efficiency which may occur from one individual to another.

GENERAL

In the present state of the technique, in vivo gamma radiation counting is subject to significant inaccuracies. For a uniform distribution of uranium in the lung tissue, counting efficiencies for persons of different sizes and builds may vary by more than a factor of two (Table IV). This variability may be increased to a factor of three if variation in source distribution (except close to the front) is included (Table V). Factors of ten are involved in the counting efficiency if concentrated sources appear close to the detector. However, the latter situation appears to be least likely since deposition should occur in the central region of the lungs where the bronchii subdivide and branch out into the lung tissue. In addition to these errors, there are other difficulties such as background radiation and skin contamination which contribute to the uncertainty and inaccuracy of the in vivo method.

CONCLUSIONS

1. The geometrically highest attainable counting efficiency for an in vivo measurement made on an average person with a nine-inch detector is about 9 to 12 percent. Actual counting efficiencies (including radiation absorption and scattering in the human tissue) range between 2 and 10 percent.
2. Maximum counting efficiencies, assuming uniform distribution of radioactivity in the lungs, are attained with the detector placed at the back of the chest. Efficiencies are 20 to 30 percent higher than with the same detector positioned at either the sides or front.
3. The heart and spinal column act as important radiation attenuators and affect the counting efficiency when making measurements at the front and back respectively.
4. Counting efficiency is sensitive to the physical dimensions of individuals. With respect to an average person with the detector at the front of the chest, the efficiency might increase 30 percent for an extremely small person and decrease by about the same amount for a very large one.

From the observed emission of radiation from lung deposits through the back, it is inferred that less variation with individual size is expected with the detector placed at the back of the chest rather than at the front.

5. Nonuniformities in the distribution of radioactivity in the lungs, particularly concentrated source regions, markedly affect the counting efficiency.
6. Two, 4-inch diameter discs of radioactivity centered at the centroids of the lungs of a manikin could be employed for estimating the counting efficiency, and hence the in vivo gamma counting instrument calibration for the dimensions of the manikin. The accuracy of this procedure would be about ± 100 percent.
7. In vivo gamma measurements of radioactive deposits in human lungs, exclusive of instrument, body contamination, and background errors are subject to inaccuracies, possibly as high as a factor of three. Under the assumption of uniform radioactivity distribution for chronic exposure and with other corrections such as that for chest thickness, the uncertainty may be reduced to a factor of two or less.

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APPENDIX A

COMPARISON OF COMPUTER SIMULATION WITH GAMMA RADIATION EXPERIMENT

A radiation absorption experiment of relatively simple geometry was carried out by R. E. Cofield. A point source of enriched uranium was located 12 inches from the detector face along the cylindrical axis (see Figure A-1). Masonite material of various thicknesses between 1/4 and 9 inches was employed for several counting runs. The data are presented in Table A-1.

The simple geometry is described mathematically, assuming (1) that the detector is directional, i e, only flux normal to the detector face is counted, and (2) that an average absorption coefficient can be used for the 50-Kev range under consideration. A differential element of the detector, see diagram, of area $2\pi r dr$ is taken. The flux from a point source, P , normal to the detector is given by

$$J = \frac{I_0}{4\pi x^2} \cdot e^{-\sigma t} \cdot \frac{c}{x} = \frac{c I_0 e^{-\sigma t}}{4\pi x^3} \quad (1)$$

Figure A-1. LOCATION OF POINT SOURCE FROM DETECTOR FACE.

Table A-1
COUNT RATE FROM ENRICHED URANIUM POINT SOURCE AS A FUNCTION OF MASONITE
SHIELDING THICKNESS

Masonite Thickness (inches)	Count Rate, I (cpm in 165-215 Kev range)	Ratio: I_0/I_t
0	4875	1.000
1/4	4411	0.905
1/2	4285	0.879
3/4	4023	0.825
1	3837	0.787
2	2947	0.605
3	2203	0.452
4	1806	0.370
5	1268	0.260
6	1005	0.206
7	790	0.162
8	592	0.121
9	424	0.087

Thus, the intensity over the differential area is

$$dI = J \cdot \frac{c}{x} \cdot 2\pi r dr \quad (2)$$

where $x^2 = r^2 + c^2$, and $t = \frac{d-e}{c} \cdot x = kx$

By substitution,

$$dI = \frac{c \cdot I_0}{2} \cdot \frac{e^{-\sigma kx}}{x^2} dx \quad (3)$$

The solution of Equation 3 is given by,

$$I = \frac{c \cdot I_0}{2} \left[-\frac{e^{-\sigma kx}}{x} - (\sigma k) \left\{ \ln x + (\sigma k) + \frac{(\sigma k)^2}{2 \cdot 2!} + \dots + \frac{(\sigma k)^n}{n \cdot n!} \right\} \right] + k_0$$

where k_0 is defined at $r = 0$, $I = 0$.

An alternative analysis of the problem is offered by the in vivo simulation program. The system was described by a three-coordinate space divided into one-inch cubes. In Table A-2 the experimental results are compared with those obtained from a direct integration and those from the computer program.

Table A-II
COMPARISON OF EXPERIMENTAL, CALCULATION, AND SIMULATION DATA

Masonite Thickness (inches)	Experimental $I_t/I_{t=0}$ (Least squares values)	Absorption Coefficient = 0.255 in^{-1}			
		Integration		Simulation	
		I_t	$I_t/I_{t=0}$	I_t	$I_t/I_{t=0}$
0	1.000	$0.03184 I_0$	1.000	$0.03064 I_0$	1.000
2	0.587	$0.01877 I_0$	0.590	$0.01809 I_0$	0.590
5	0.261	$0.00847 I_0$	0.266	$0.00822 I_0$	0.269
9	0.091	$0.00289 I_0$	0.091	$0.00287 I_0$	0.094

It must be realized that the area of the detector for the simulation was 61 square inches; whereas, for the integration method, the area was 63.6 square inches. The maximum deviation between the integration results is about four percent. The disagreement between the experimental data and the integration results is in the average attenuation coefficient used. A trial and error solution of Equation 3 showed that under the assumptions mentioned earlier, the average attenuation coefficient in the 165 to 215 Kev energy range is about 0.255 in^{-1} , or 0.100 cm^{-1} .

APPENDIX B

GRID SIZE AND ACCURACY OF THE COMPUTER SIMULATION

This study was made using a three-coordinate space composed of one-inch cubes. In the limit as the grid size is reduced, the simulation attains complete computational accuracy, but at the expense of infinitely many more computations and greater cost. As a check on the one-inch grid size, computations performed for the average person were duplicated after obtaining data for a 1/2-inch grid (one-half-inch cubes). The latter had eight cubes to one in the larger system.

A brief comparison of counting efficiencies obtained for identical source volumes with two-coordinate systems is presented in Table B-1. It is apparent that the one-inch grid yielded efficiencies which were consistently 10 - 20 percent higher than with the smaller grid size.

Table B-1

COMPARISON OF COUNTING EFFICIENCIES DETERMINED USING ONE-INCH AND ONE-HALF-INCH GRIDS

Source Geometry	Grid Size (in)	Counting Efficiency	
		Zero Coefficient	Linear Absorption Coefficient
Front Half of Chest	1	0.147	0.060
	1/2	0.128	0.052
Center of Right Lung (one cubic inch)	1	0.149	0.065
	1/2	0.127	0.055
Front of Right Lung (one cubic inch)	1	0.135	0.269
	1/2	0.126	0.259

Larger discrepancies are expected at the periphery of the lung tissue where the smaller grid is capable of producing a more accurate description of the lung tissue. Furthermore, a volume is only considered to contain a source if more than 50 percent of the cube contains lung tissue.

APPENDIX C

DESCRIPTION OF COMPUTER PROGRAM

The Fortran program is included in Table C-1, a simplified flowsheet is shown in Figure C-1. The following subsections deal with the mechanics of input and output for the program.

INPUT DATA

The input data is read into the computer from Logical Tape 10 in the following order:

Format 1 of Table C-II contains the desired values of attenuation coefficients. Five sets may be supplied to the program with each set consisting of the four coefficients for the four types of body tissue being considered. The five sets of coefficients of the example in Format 1 are absorption, total linear, zero, average absorption, and average total linear; each set contains the body tissues: soft tissue, lung, bone, and heart.

Format 2 contains the detector size and position for the front, right side, back, and left side of the chest. The coordinates are based on the grid represented in Figure C-1. Each coordinate is a two-digit number. The first four numbers give the sides of the detector; the next two numbers are the bottom and top edges of the detector, and the last number is the desired radius. The example in Format 2 has the following data:

<u>Column</u>	<u>Value</u>	<u>Meaning</u>
1 - 2	6	Detector on front. Left side at X = 6.
3 - 4	14	Detector on front. Right side at X = 14.
5 - 6	2	Detector on right. Right side at Y = 2.
7 - 8	10	Detector on right. Left side at Y = 10.
9 - 10	6	Detector on back. Right side at X = 6.
11 - 12	14	Detector on back. Left side at X = 14.
13 - 14	2	Detector on left. Left side at Y = 2.
15 - 16	10	Detector on left. Right side at Y = 10.
17 - 18	1	Bottom of detector at Z = 1.
19 - 20	9	Top of detector at Z = 9.
21 - 24	4.25	Detector radius is 4.25 units.

Format 3 tells which of the four detector positions are to be used and the size of the chest cavity to be considered. The first two numbers are the first and last positions of the detector, where 1 means front, 2 - right side, 3 - back side, and 4 - left side. All intermediate positions will be used. The example data has the detector on the right side, the back, and then the left side. The next number is one-seventh of the maximum "X" value, which must be an integral multiple of seven. The last two numbers are the maximum "Y" and "Z" values, respectively.

Table C-1

FORTRAN PROGRAM FOR IN VIVO SIMULATION

```

C    EFFECT OF VARIABLES ON IN-VIVO RADIATION DETECTION
      DIMENSION IUGBA(28,15,20),TD(28,20),CUS(5),CUG(5),CBO(5),CAR(5),
      ISUMT(5),SSUMT(5),XSUMT(28),YSUMT(28),ZSUMT(28),U(5),G(5),B(5),
      2A(5),IUS(7),IUG(7),IBO(7),IAR(7),T(5),TD2(28,20),TD3(28,20)
      READINPUTTAPE10,7,(CUS(1),CUG(1),CBO(1),CAR(1),I#1,5)
      READ INPUT TAPE10,4,IDB1,IDE1,IDB2,IDE2,IDB3,IDE3,IDB4,IDE4,
      IDBOT,IDTOP,DETRAD
      3 FORMAT(I3,3I2,6(I4,3I2))
      4 FORMAT(10I2,F6.2)
      7 FORMAT(10F7.4)
      READ INPUT TAPE 10,4,ITBE,ITEN,MZ,JZ,KZ
      DO 5 K#1,KZ
      DO 5 J#1,JZ
      DO 5 MM#1,MZ
      READINPUTTAPE10,3,(IUS(LL),IUG(LL),IBO(LL),IAR(LL),LL#1,7)
      DO 5 LL#1,7
      NI#(MM-1)*7+LL+28*(J-1)+420*(K-1)
      CALL PACK(IUGBA,LL,IUS,IUG,IBO,IAR,NI)
      5 CONTINUE
      READINPUTTAPE10,4,IIA,IIB,IIC,IID
      CALL EFMI
      PI#12.56637
      DO 534IT#ITBE,ITEN
      GOTO(70,71,70,71),IT
      70 IAA#IIA
      IBB#IIB
      GOTO 178
      71 IAA#IIC
      IBB#IID
      178 DO 206 N#1,20
      DO 206 L#1,28
      TD2(L,N)#0.0
      TD3(L,N)#0.0
      206 TD(L,N)#0.0
      72 JOAN#0
      JILL#0
      DO 42 MN#1,5
      42 SSUMT(MN)#0.0
      DO 52 J#1,28
      52 YSUMT(J)#0.0
      DO 53 K#1,28
      53 ZSUMT(K)#0.0
      DO 54 I#1,28
      54 XSUMT(I)#0.0
      DO 200 I#IIA,IBB
      SX#I
      PAUL#SOSF(5)
      DO 200 J#1,JZ
      SY#J
      DO 200 K#1,KZ
      IF(IUGBA(I,J,K))200,200,15

```

Table C-1 (Continued)

```

15 SZ#K
DO 16 MN#1,5
16 SUMT(MN)#0.0
GOTO(150,151,153,154),IT
150 M#1
DY#1.0
IA#IDB1
IB#IDE1
GOTO201
151 L#IIC
DX#L
IA#IDB2
IB#IDE2
GOTO201
153 M#JZ
DY#JZ
IA#IDB3
IB#IDE3
GOTO201
154 L#IID
DX#L
IA#IDB4
IB#IDE4
201 DO 175 IAB#IA,IB
DO 175 N#IDBOT,IDTOP
DZ#N
GOTO(202,203,202,203),IT
202 DX#IAB
L#IAB
DET#DX
GO TO 25
203 DY#IAB
M#IAB
DET#DY
25 IF(((DET-FLOATF((IA+IB)/2))**2)+((DZ-FLOATF((IDBOT+IDTOP)/2))**2)
1-(DETRAD**2))309,309,175
309 SUMUS#0.0
SUMUG#0.0
SUMBO#0.0
SUMAR#0.0
R#SQRTF((SX-DX)*(SX-DX)+(SY-DY)*(SY-DY)+(SZ-DZ)*(SZ-DZ))
IF(XABSF(I-L)-XABSF(J-M))30,30,35
30 IF(XABSF(J-M)-XABSF(K-N))44,44,46
35 IF(XABSF(I-L)-XABSF(K-N))44,44,47
47 ICC#XMINOF(I,L)
IDD#XMAXOF(I,L)
AP#R/ABSF(SX-DX)
NN#-1
GOTO41
46 ICC#XMINOF(J,M)
IDD#XMAXOF(J,M)
AP#R/ABSF(SY-DY)

```

Table C-1 (Continued)

```

NN#0
GOTO41
44 ICC#XMINOF(K,N)
   IDD#XMAXOF(K,N)
   AP#R/ABSF(SZ-DZ)
NN#1
41 IF DIVIDE CHECK 305,306
305 WRITE OUTPUT TAPE9,321,NN,AP,SX,DX,SY,DY,SZ,DZ
306 DO 65 ICD#ICC,IDD
   IF(NN)55,56,57
55 IF(I-L)58,58,59
56 IF(J-M)58,58,59
57 IF(K-N)58,58,59
58 CD#ICD
   GO TO160
59 CD#IDD+ICC-ICD
160 IF(NN)24,23,22
22 KSSZ# CD
   ISSX#( CD-DZ)*(SX-DX)/(SZ-DZ)+DX+0.5
   JSSY#( CD-DZ)*(SY-DY)/(SZ-DZ)+DY+0.5
   GOTO26
23 JSSY# CD
   KSSZ#( CD-DY)*(SZ-DZ)/(SY-DY)+DZ+0.5
   ISSX#( CD-DY)*(SX-DX)/(SY-DY)+DX+0.5
   GOTO26
24 ISSX# CD
   JSSY#( CD-DX)*(SY-DY)/(SX-DX)+DY+0.5
   KSSZ#( CD-DX)*(SZ-DZ)/(SX-DX)+DZ+0.5
26 IF DIVIDE CHECK 307,308
307 WRITE OUTPUT TAPE9,321,NN,AP,SX,DX,SY,DY,SZ,DZ
308 N2#ISSX+28*(JSSY-1)+420*(KSSZ-1)
   CALL UNPACK(N2,IUGBA,IJS,UG,BO,AR)
   IF(NN)216,217,218
218 IF(KSSZ-K)220,219,220
217 IF(JSSY-J)220,219,220
216 IF(ISSX-I)220,219,220
219 US#US/2.0
   UG#UG/2.0
   BO#BO/2.0
   AR#AR/2.0
220 SUMUS#SUMUS+US
   SUMUG#SUMUG+UG
   SUMBO#SUMBO+BO
   SUMAR#SUMAR+AR
65 CONTINUE
   DO 174 MN#1,5
   U(MN)#CUS(MN)*SUMUS
   G(MN)#CUG(MN)*SUMUG
   B(MN)#CBO(MN)*SUMBO
   A(MN)#CAR(MN)*SUMAR
   ATEN #-(U(MN)+G(MN)+B(MN)+A(MN))*AP
   GOTO(155,156,155,156),IT

```

Table C-I (Continued)

```

155 T(MN)#(EXP(ATEX))/ (PI*R*R) *ABS((SY-DY)/R)
    INDEX#L
    GOTO158
156 T(MN)#(EXP(ATEX))/ (PI*R*R) *ABS((SX-DX)/R)
    INDEX#M
158 SUMT(MN)#SUMT(MN)+T(MN)
174 CONTINUE
    TD(INDEX,N)#TD(INDEX,N)+T(1)
    TD2(INDEX,N)#TD2(INDEX,N)+T(2)
    TD3(INDEX,N)#TD3(INDEX,N)+T(3)
175 CONTINUE
    XSUMT(I)#XSUMT(I)+SUMT(I)
    YSUMT(J)#YSUMT(J)+SUMT(I)
    ZSUMT(K)#ZSUMT(K)+SUMT(I)
    IF(JOAN)181,6,181
    6 GOTO(350,351,353,354),IT
350 WRITEOUTPUTTAPE9,313
    GOTO205
351 WRITEOUTPUTTAPE9,318
    GOTO205
353 WRITEOUTPUTTAPE9,314
    GOTO205
354 WRITEOUTPUTTAPE9,319
205 WRITEOUTPUTTAPE9,300
    WRITEOUTPUTTAPE9,301
    WRITEOUTPUTTAPE9,302
181 WRITEOUTPUTTAPE9,303,K,J,I,(SUMT(MN),MN#1,5)
    JOAN#JOAN+1
    IF(JOAN-45)185,184,185
184 JOAN#0
185 DO 187 MN#1,5
187 SSUMT(MN)#SSUMT(MN)+SUMT(MN)
200 CONTINUE
    GOTO(290,291,292,293),IT
290 WRITEOUTPUTTAPE9,313
    GOTO294
291 WRITEOUTPUTTAPE9,318
    GOTO294
292 WRITEOUTPUTTAPE9,314
    GOTO294
293 WRITEOUTPUTTAPE9,319
294 WRITEOUTPUTTAPE9,300
    WRITEOUTPUTTAPE9,304,(SSUMT(MN),MN#1,5)
    WRITEOUTPUTTAPE9,310
    WRITEOUTPUTTAPE9,311
    DO 523 MM#1,28
523 WRITEOUTPUTTAPE9,312,(MM,XSUMT(MM),MM,YSUMT(MM),MM,ZSUMT(MM))
    GOTO(530,532,530,532),IT
530 DO 531 L#IA,IB
    DO 531 N#IDBOT,IDTOP
    IF(JILL)230,231,230
231 WRITEOUTPUTTAPE9,315

```

Table C-I (Continued)

```

232 WRITEOUTPUTTAPE9,316
230 JILL#JILL+1
    IF(JILL-45)531,271,531
271 JILL#0
531 WRITEOUTPUTTAPE9,317,L,N,TD(L,N) ,TD2(L,N),TD3(L,N)
    GOTO2005
532 DO533M#1A,IB
    DO533N#IDBOT,IDTOP
    IF(JILL)331,330,331
330 WRITEOUTPUTTAPE9,315
233 WRITEOUTPUTTAPE9,320
331 JILL#JILL+1
    IF(JILL-45)533,273,533
273 JILL#0
533 WRITEOUTPUTTAPE9,317,M,N,TD(M,N) ,TD2(M,N),TD3(M,N)
2005 PAUL#SOSF(5)
534 CONTINUE
2006 CALL EXIT
300 FORMAT(116H0          BASIS--          ABS. C.          SCATT
    I. C.          C.# 0          AVG. ABS. C.          AVG. SCATT. C.)
301 FORMAT(24H  INTENSITY OF + POINTS)
302 FORMAT(20H      K      J      I)
303 FORMAT(I6,2I7,E24.8,4E18.8)
304 FORMAT(22H0      TOTAL  INTENSITY,E22.8,4E18.8)
310 FORMAT(73H0      SUM  INTENSITIES  BY  PLANES,  USING  ABSORPTIC
    IN  COEFFICIENTS-)
311 FORMAT(95H0      I      INTENSITY          J      INTEN
    ISITY          K      INTENSITY)
312 FORMAT(I9,E20.8,2(I15,E20.8))
313 FORMAT(33H1  DETECTOR  ON  FRONT          )
314 FORMAT(34H1  DETECTOR  ON  BACK          )
315 FORMAT(47H1  INTENSITY  SUMMED  BY  DETECTOR  POINTS )
316 FORMAT(46H0      DETECTOR  POINT  L      N      INTENSITY)
317 FORMAT(I25,I5,3E19.8)
318 FORMAT(32H1  DETECTOR  ON  RIGHT  SIDE)
319 FORMAT(31H1  DETECTOR  ON  LEFT  SIDE)
320 FORMAT(46H0      DETECTOR  POINT  M      N      INTENSITY)
321 FORMAT(4H D-C I4,7F7.2)
    END(0,1,0,0,0)
SUBROUTINE UNPACK(N2,IUGBA,US,UG,BO,AR)
C  TO UNPACK DATA FROM IN-VIVO INPUT
    DIMENSION IUGBA(10)
S  STZ IUUS
S  STZ IUUG
S  STZ IBBO
S  STZ IAAR
S  CLA IUUS
S  ADM IUGBA(N2)
S  LRS 12
S  ALS 18
S  STO IUUS
S  ARS 25

```


Table C-1 (Continued)

```

S      LLS 4
S      ALS 18
S      STO IUUG
S      ARS 25
S      LLS 4
S      ALS 18
S      STO IBBO
S      ARS 25
S      LLS 4
S      ALS 18
S      STO IAAR
      US#FLOATF(IUUS)/10.0
      UG#GLOATF(IUUG)/10.0
      BO#FLOATF(IBBO)/10.0
      AR#FLOATF(IAAR)/10.0
      RETURN
      END(0,1,0,0,0)
SUBROUTINE PACK(IUGBA,LL,IUS,IUG,IBO,IAR,NI)
C      TO PACK DATA FOR IN-VIVO INPUT
      DIMENSION IUGBA(10),IUS(7),IUG(7),IBO(7),IAR(7)
S      CLA IAR(LL)
S      ARS 4
S      ADD IBO(LL)
S      ARS 4
S      ADD IUG(LL)
S      ARS 4
S      ADM IUS(LL)
S      ARS 6
S      STO IUGBA(NI)
      IF(IUG(LL)-5)81,80,80
S 80 CLA IUGBA(NI)
S      SSP
S      STO IUGBA(NI)
      GO TO 5
S 81 CLA IUGBA(NI)
S      SSM
S      STO IUGBA(NI)
S      5 RETURN
      END(0,1,0,0,0)

```

Format 4 shows a typical card containing the fractional part of each tissue at the various grid points. Each card contains seven grid points. The field for each grid point is ten columns wide. Each field has Columns one and ten blank. The remaining columns give the fractions of soft tissue, lung, bone, and heart in the cube around that grid point. Consider the first field of Format 4.

Figure C-1. FLOW DIAGRAM OF COMPUTER PROGRAM.

The remaining 0.2 is air. The cube described by Field 2 is all soft tissue. The cubes represented by Fields 3 and 4 are considered as source points, since the fraction of lung is greater than, or equal to, one-half. The number of cards of the Format 4 type will be the product:

$$(1/7) \times (\text{maximum X value}) \times (\text{maximum Y value}) \times (\text{maximum Z value}).$$

When the detector is placed on the right or left side, the arms may be raised to get a better count. This reduces the range of "X". Format 5 contains this information. In this example, when the detector is on the front or back, "X" ranges from 1 to 24. When the detector is on the right side its face is in the plane X=4, and on the left side its face is on the plane X=20. Only the values of "X" between 4 and 20 are then considered as body points by the program.

Table C-II
INPUT DATA FORMATS

Format	Column Number															
	1	5	10	15	20	25	30	35	40	45	50	55	60	65	70	
1		.0762	.0152	.1544	.0762		.3607	.0721	.6248		.3607	.0000		.0000		
		.0000	.0000	.0508	.0508		.0508	.0508	.2540		.2540	.2540		.2540		
2	0614021006	1402100109	4.25													
3	0204031111															
4	03020300	10000000	02050003	02080000	00000109	01020304	02030050									
5	01240420															

OUTPUT DATA

The output is all placed on Logical Tape 9 to be printed off line. The output includes:

1. Every source point and its corresponding I/I_0 values for the five sets of attenuation coefficients, Table C-III.
2. The sum of the I/I_0 values for all source points and for each type of tissue for the five sets of attenuation coefficients, Table C-IV.
3. The I/I_0 value attenuated in each X, Y, and Z plane, Table C-IV.
4. The I/I_0 value received at each detector point from all source points for the first three sets of attenuation coefficients, Table C-V.

Table C-III

SAMPLE COMPUTER OUTPUT LIST OF THE ATTENUATIONS FOR EACH SOURCE POINT

DETECTOR	ON	FRONT					
BASIS--			ABS. C.	SCATT. C.	C.= 0	AVG. ABS. C.	AVG. SCATT. C.
INTENSITY OF + POINTS				(LINEAR)			(LINEAR)
K (Z)	J (Y)	I (X)					
4	3	8	0.22742633E-00	0.13643191E-00	0.26230521E-00	0.23263808E-00	0.14610785E-00
5	3	8	0.23130268E-00	0.13482232E-00	0.26940615E-00	0.24036148E-00	0.15462951E-00
3	4	8	0.15039033E-00	0.87046022E-01	0.17585385E-00	0.14779933E-00	0.75170687E-01
4	4	8	0.16436020E-00	0.94373160E-01	0.19229707E-00	0.16253719E-00	0.84320489E-01
5	4	8	0.16710611E-00	0.91856499E-01	0.19779690E-00	0.16804785E-00	0.89104513E-01
6	4	8	0.16194674E-00	0.89183862E-01	0.19229707E-00	0.16484687E-00	0.91497691E-01
3	5	8	0.10821389E-00	0.52681901E-01	0.13350531E-00	0.10670729E-00	0.44288606E-01
4	5	8	0.11921491E-00	0.61321168E-01	0.14467815E-00	0.11642053E-00	0.49588010E-01
5	5	8	0.12325172E-00	0.64546189E-01	0.14851084E-00	0.12052947E-00	0.53289982E-01
6	5	8	0.11848352E-00	0.59103769E-01	0.14467815E-00	0.11743841E-00	0.52265569E-01
7	5	8	0.10520925E-00	0.45752874E-01	0.13350531E-00	0.10844589E-00	0.48868663E-01
8	5	8	0.86561137E-01	0.31913301E-01	0.11616455E-00	0.94562320E-01	0.43551949E-01
3	6	8	0.80891085E-01	0.34512486E-01	0.10403488E-00	0.79354121E-01	0.27319093E-01
4	6	8	0.89086194E-01	0.41745197E-01	0.11153496E-00	0.85676184E-01	0.30303126E-01
5	6	8	0.91763464E-01	0.43887099E-01	0.11413332E-00	0.88373017E-01	0.32390326E-01
6	6	8	0.88940085E-01	0.41145426E-01	0.11153496E-00	0.86333840E-01	0.31774940E-01
7	6	8	0.81204416E-01	0.34672558E-01	0.10403488E-00	0.80392633E-01	0.29629467E-01
8	6	8	0.69346973E-01	0.25403690E-01	0.92549192E-01	0.71120562E-01	0.25911205E-01
3	7	8	0.62460765E-01	0.24263120E-01	0.82795102E-01	0.60178001E-01	0.17044561E-01
4	7	8	0.67730894E-01	0.28432135E-01	0.87872510E-01	0.64550474E-01	0.19074994E-01
5	7	8	0.68273903E-01	0.27809553E-01	0.89635395E-01	0.65978359E-01	0.19707505E-01
6	7	8	0.66758633E-01	0.26908036E-01	0.87872510E-01	0.64769023E-01	0.19564468E-01
7	7	8	0.62351537E-01	0.24049029E-01	0.82795102E-01	0.60806144E-01	0.18238305E-01
8	7	8	0.55440120E-01	0.19574208E-01	0.75023148E-01	0.54740680E-01	0.16155270E-01
3	8	8	0.49442342E-01	0.17624741E-01	0.67083327E-01	0.46460938E-01	0.10832435E-01
4	8	8	0.53062075E-01	0.20416988E-01	0.70575757E-01	0.49387266E-01	0.12002677E-01
5	8	8	0.53345014E-01	0.20005986E-01	0.71786654E-01	0.50322205E-01	0.12345366E-01
6	8	8	0.52058496E-01	0.19015063E-01	0.70575757E-01	0.49518919E-01	0.12258702E-01
7	8	8	0.49618506E-01	0.18004812E-01	0.67083329E-01	0.46875715E-01	0.11493533E-01
2	9	8	0.33963661E-01	0.84829613E-02	0.51421521E-01	0.33610910E-01	0.62203242E-02
3	9	8	0.39606585E-01	0.12890438E-01	0.55211551E-01	0.36585927E-01	0.71361527E-02
4	9	8	0.41568577E-01	0.13946327E-01	0.57658864E-01	0.38392201E-01	0.76392139E-02
5	9	8	0.42004675E-01	0.13975148E-01	0.58504955E-01	0.39103769E-01	0.79212740E-02
6	9	8	0.40838365E-01	0.12980644E-01	0.57658864E-01	0.38518196E-01	0.78205884E-02
1	10	8	0.22920925E-01	0.35378090E-02	0.39890967E-01	0.24547871E-01	0.35789604E-02
2	10	8	0.28146067E-01	0.66917926E-02	0.43348607E-01	0.27033634E-01	0.41461518E-02
3	10	8	0.31935547E-01	0.92576398E-02	0.46075428E-01	0.29114092E-01	0.46958689E-02
4	10	8	0.33314697E-01	0.98171324E-02	0.47823590E-01	0.30353185E-01	0.49861621E-02
3	3	9	0.22625153E-00	0.13438711E-00	0.26230521E-00	0.23103550E-00	0.14092711E-00
4	3	9	0.24322500E-00	0.14396871E-00	0.28175458E-00	0.24910516E-00	0.15392834E-00
5	3	9	0.24925288E-00	0.14961787E-00	0.28771785E-00	0.25595798E-00	0.16234578E-00
6	3	9	0.24536958E-00	0.13530132E-00	0.28175458E-00	0.25507969E-00	0.17544726E-00
3	4	9	0.15851165E-00	0.80283792E-01	0.19229707E-00	0.16122344E-00	0.80559546E-01
4	4	9	0.17263627E-00	0.89131585E-01	0.20826082E-00	0.17555221E-00	0.89429850E-01
5	4	9	0.17811778E-00	0.94490810E-01	0.21348564E-00	0.18100695E-00	0.94648693E-01

Table C-IV

SAMPLE COMPUTER OUTPUT LIST OF THE SUMMED ATTENUATIONS FOR SOURCE POINTS

DETECTOR	CN	FRONT				
BASIS-		ABS. C.	SCATT. C. (LINEAR)	C.#0	AVG. ABS. C.	AVG. SCATT. C. (LINEAR)
TOTAL INTENSITY		0.17646112E 02	0.72493747E 01	0.23284813E 02	0.17949696E 02	0.70562160E 01
FRACTIONAL ATTENUATION-						
MUSCLE		0.14275231E-00	0.47537991E-00	0.	0.98197437E-01	0.36435977E-00
LUNG		0.26672497E-01	0.66606863E-01	0.	0.87975588E-01	0.22223904E-01
BONE		0.25895955E-01	0.55878004E-01	0.	0.89230814E-02	7.27540614E-01
HEART		0.49585611E-01	0.94790845E-01	0.	0.34019615E-01	0.82821153E-01
SUM INTENSITIES BY PLANES, USING ABSORPTION COEFFICIENTS-						
I(X)	INTENSITY	J(Y)	INTENSITY	K(Z)	INTENSITY	
1	0.	1	0.	1	0.24626744E-01	
2	0.	2	0.	2	0.33431939E-00	
3	0.	3	0.22390158E 01	3	0.33055800E 01	
4	0.	4	0.29833312E 01	4	0.34795178E 01	
5	0.	5	0.31793499E 01	5	0.38717759E 01	
6	0.94283949E 00	6	0.30799945E 01	6	0.33091484E 01	
7	0.24608418E 01	7	0.28547888E 01	7	0.17956180E 01	
8	0.36540002E 01	8	0.20740206E 01	8	0.12562833E 01	
9	0.41615271E 01	9	0.10330003E 01	9	0.25676236E-00	
10	0.10036207E 01	10	0.19896212E-00	10	0.	
11	0.	11	0.	11	0.	
12	0.65223223E 00	12	0.	12	0.	
13	0.17052589E 01	13	0.	13	0.	
14	0.15639406E 01	14	0.	14	0.	
15	0.10932283E 01	15	0.	15	0.	
16	0.39614689E-00	16	0.	16	0.	
17	0.	17	0.	17	0.	
18	0.	18	0.	18	0.	
19	0.	19	0.	19	0.	
20	0.	20	0.	20	0.	
21	0.	21	0.	21	0.	

Table C-V

SAMPLE COMPUTER OUTPUT LIST OF THE ATTENUATIONS FOR EACH DETECTOR POINT

INTENSITY	SUMMED	BY	DETECTOR	POINTS		
DETECTOR	POINT	L(X)	N(Z)	INTENSITY		
6	1		0.	0.	0.	0.
6	2		0.	0.	0.	0.
6	3		0.	0.	0.	0.
6	4		0.28328376E-00	0.14737200E-00	0.34742678E-00	0.34742678E-00
6	5		0.27925828E-00	0.13761380E-00	0.34742618E-00	0.34742618E-00
6	6		0.25632744E-00	0.11841986E-00	0.32435956E-00	0.32435956E-00
6	7		0.	0.	0.	0.
6	8		0.	0.	0.	0.
6	9		0.	0.	0.	0.
7	1		0.	0.	0.	0.
7	2		0.25378060E-00	0.12506496E-00	0.31909224E-00	0.31909224E-00
7	3		0.31949309E-00	0.16716595E-00	0.39220499E-00	0.39220499E-00
7	4		0.34793337E-00	0.17505496E-00	0.43033617E-00	0.43033617E-00
7	5		0.34158669E-00	0.16285036E-00	0.42861678E-00	0.42861678E-00
7	6		0.30853763E-00	0.13726519E-00	0.39425346E-00	0.39425346E-00
7	7		0.27162018E-00	0.12821650E-00	0.34152614E-00	0.34152614E-00
7	8		0.22137264E-00	0.97441057E-01	0.28480097E-00	0.28480097E-00
7	9		0.	0.	0.	0.
8	1		0.	0.	0.	0.
8	2		0.26986735E-00	0.12203991E-00	0.34747834E-00	0.34747834E-00
8	3		0.35019263E-00	0.17396905E-00	0.43596392E-00	0.43596392E-00
8	4		0.39097567E-00	0.19098305E-00	0.48774616E-00	0.48774616E-00
8	5		0.38782998E-00	0.18132775E-00	0.48958042E-00	0.48958042E-00
8	6		0.35268750E-00	0.16074564E-00	0.44797190E-00	0.44797190E-00
8	7		0.30599122E-00	0.14596619E-00	0.38393326E-00	0.38393326E-00
8	8		0.24552609E-00	0.10641830E-00	0.31731222E-00	0.31731222E-00
8	9		0.	0.	0.	0.
9	1		0.18692221E-00	0.60054096E-01	0.26481551E-00	0.26481551E-00
9	2		0.25799320E-00	0.10294690E-00	0.34502374E-00	0.34502374E-00
9	3		0.33986551E-00	0.16145171E-00	0.43022280E-00	0.43022280E-00
9	4		0.38298919E-00	0.17797821E-00	0.48543803E-00	0.48543803E-00
9	5		0.39090763E-00	0.18221321E-00	0.49510635E-00	0.49510635E-00
9	6		0.36301058E-00	0.16644195E-00	0.46137373E-00	0.46137373E-00
9	7		0.31744709E-00	0.15036120E-00	0.39958727E-00	0.39958727E-00
9	8		0.25562649E-00	0.10765830E-00	0.33251204E-00	0.33251204E-00
9	9		0.20157599E-00	0.75453629E-01	0.27010809E-00	0.27010809E-00
10	1		0.17202754E-00	0.46189041E-01	0.25418381E-00	0.25418381E-00
10	2		0.22774143E-00	0.75302275E-01	0.32027427E-00	0.32027427E-00
10	3		0.28733585E-00	0.11009336E-00	0.38763880E-00	0.38763880E-00
10	4		0.32917139E-00	0.13251551E-00	0.43549084E-00	0.43549084E-00
10	5		0.34678833E-00	0.14812804E-00	0.45118373E-00	0.45118373E-00
10	6		0.33243063E-00	0.13951658E-00	0.43256214E-00	0.43256214E-00
10	7		0.30095049E-00	0.13172079E-00	0.38713049E-00	0.38713049E-00
10	8		0.25052010E-00	0.99860873E-01	0.33029564E-00	0.33029564E-00
10	9		0.20197323E-00	0.73599210E-01	0.27240960E-00	0.27240960E-00

APPENDIX D

ILLUSTRATION OF COMPUTER CALCULATION

The computer computation entails determining the path of radiation between a source point, S, and a detector point, D. The type and location of intermediate tissues are determined from this path. Using the length of tissue through which radiation passes along with the appropriate attenuation coefficient (Table D-1), the fractional reduction in intensity due to absorption and scattering may be calculated.

Figure D-1 shows a source, S, and a detector point, D, located in an X-Y plane. In this example, the distance between points "S" and "D" is given by:

$$\omega = \sqrt{(5-1)^2 + (4-7)^2} = 5 \text{ inches} \quad (1)$$

The cubes nearest to the line SD, such as represented by points b, d, and e, are used to determine the composition of tissues through which the radiation passes. The attenuation distance, l_i , of tissue-type "i" in a cube is given by:

$$l_i = f_i \frac{\omega}{\Delta x} (1 \text{ inch}) = f_i \frac{\sqrt{(5-1)^2 + (4-7)^2}}{(5-1)} (1 \text{ inch}) \quad (2)$$

where f_i is the fraction of the ith tissue in the volume. After estimating the l_i for each tissue type and each cube along the path SD, the radiation flux I , at "D" is calculated by:

$$I = I_0 \frac{e^{-\sum_i \sigma_i l_i}}{4 \pi \omega^2} \quad (3)$$

where I_i equals $\sum_i I_i$. The flux is given in units of photons per unit time per unit area. The tissue attenuation coefficients are given in Table D-1.

The total radiation incident upon the detector at Point D is computed by multiplying the flux (I) by the projected area of the detector which is normal to the flux. This projection is represented by PD in Figure D-1 and is equal to the unit area times the cosine of the Angle A shown. Thus,

$$J = \frac{I_0 e^{-\sum_i \sigma_i l_i}}{4 \pi \omega^2} A_p \quad (4)$$

where:

$$A_p = A \frac{(5-1)}{\sqrt{(5-1)^2 + (4-7)^2}} = A \frac{4}{5} = 0.80A \quad (5)$$

Table D-I
ATTENUATION COEFFICIENTS USED FOR HUMAN TISSUES AT 186 Kev

Tissue	Linear Absorption Coefficient - $\sigma_a + f\sigma_s$ (in^{-1})	Absorption Coefficient - σ_a (in^{-1})
Lung	0.0724	0.0152
Heart	0.361	0.0762
Muscle and Fats	0.361	0.0762
Bone	0.625	0.1544

Figure D-1. PATH OF RADIATION FOR A GIVEN SOURCE POINT AND DETECTOR POINT.