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Geometrical and Attenuation Corrections for a Two Crystal Whole-Body Counter; Body Scanning with An 8 x 4-inch Sodium Iodide Crystal.

By

Gerald J. Hine, Sebastian Genna and Belton A. Burrows

From Boston University School of Medicine and the Radioisotope and Medical Services, Veterans Administration Hospital, Boston, Mass.

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INTRODUCTION

Criteria for the design of a whole-body counter for studying the metabolism of administered radioisotopes in humans have been well stated by many investigators. From a discussion by Miller (1) we may quote as follows: "If reliable and reproducible total-body measurements are to be made, a counter should be designed to satisfy the following specifications, listed in the order of importance.

- (1) The sensitivity of the counter should not depend upon the subject's body build or upon the distribution of the material within the body.
- (2) It should be possible to calibrate the counter with reasonable accuracy by using an appropriate phantom.
- (3) The position of the subject and of each crystal should be the same for all total-body measurements, regardless of the subject's body build."

Numerous whole-body counters using one or several sodium iodide crystals of various sizes have been described (1). Though for some installations additional factors such as available space or economical

(1) Whole-Body Counting, Proceedings of a Symposium, Vienna 1961, International Atomic Energy Agency, Vienna (1962).

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conditions were of importance, the above stated criteria were of concern to most of the authors. However, the gamma-ray absorption for each individual patient has not been evaluated to obtain results which do not depend on the subject's body build and also upon the distribution of the radioisotope within the body.

In Table I the gamma-rays of various energies are listed which cover the spectrum of radioisotopes most commonly used in medical research. Assuming an original intensity of 100% at a given body area, that fraction of the gamma-rays that is lost by their interaction with tissue of a given thickness is stated in Table I. For example, the 606 keV gamma-rays of I^{131} diminish to about one-half (46%) of their initial abundance when penetrating 5 cm of soft tissue. Therefore, from any I^{131} labeled compound distributed evenly within the body, or concentrated mainly within one organ, only a small fraction of the I^{131} gamma-rays will escape from the body without an interaction.

The present paper describes a method which allows a determination of the average gamma-ray absorption for each individual body count. No assumption has to be made concerning the radioisotope distribution within the body. The actual distribution can remain unknown.

In addition to the body counts it is frequently desirable to plot the radioisotope distribution by a body scan. In our installation we

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one of our 8 x 4-in. sodium iodide crystals from the body counter for body scanning within the iron room. Since a focussing collimator for an 8-in. diameter crystal has not been reported previously, its construction and properties will be discussed briefly.

WHOLE-BODY COUNTER

Iron room and crystal mounting.

The iron room, located in an annex of the basement of the Boston University Medical School Research Building, has been built of 9-in. thick armor plates fabricated during World War II. In order to provide maximum access to the room, two sliding doors each 3 ft. wide, close one of the 12 ft. walls of the room (Fig. 1). The doors can be operated independently, either manually or by motor drive.

The two 8 x 4-in. sodium iodide crystals^{*} are mounted on tracks along the back wall of the iron room (Fig. 1). Each crystal can be raised, lowered and tilted with respect to the horizontal plane. Furthermore, they travel the length of the steel room along its center line, since the crystals are mounted at a fixed distance from the back wall of the steel room. This keeps the two crystals permanently aligned along the long axis of the patient who is positioned on a plane surface.

With a two crystal array moderate sensitivity was obtained without undue sacrifice of uniformity in counting efficiency. Figure 2 shows the

^{*} Harshaw Chemical Company, Cleveland, Ohio

results for a 100 cm separation of the centers of the two crystals. The best results were obtained by placing the crystals at a 45° angle at 57 cm above a line which should correspond to the long axis of a human subject. For sources along this center line in air, reasonably uniform counting efficiency is obtained over a 100 cm long center section. The supra-sternal notch of each subject is positioned beneath the center of the left crystal. The knee of a subject of average height comes then at a point somewhat beyond the center of the right crystal (Fig. 2). The support on which the subject lies is at 67 cm from the crystal center plane.

Additional variations in counting efficiency will occur for points displaced above, below or lateral to the center line. For a 13 cm lateral displacement the mean efficiency falls by 3%. Considerable variations for positions above and below the center line can occur, particularly for points located directly below one of the crystals. If, however, the subject is measured in both the supine and prone position, these variations are markedly reduced. The mean value of the supine plus prone determinations can be demonstrated to vary by less than $\pm 7\%$ over a region 10 cm above or below the center line.

Correction for gamma-ray attenuation within the human body.

The counting efficiency for a given amount of radioactivity will not only vary with changes in geometrical location relative to the two crystals.

but also with differences in tissue thickness between the source and the body surface. One way to minimize variations in gamma-ray absorption is to have all measurements done in pairs by adding the count-rates in the supine and prone position. Then a source of radioactivity close to the skin would have in one position a minimum of tissue as absorber and a maximum in the other position. Furthermore, a source which is in one position closer to the crystals than the center line, will be further away in the other position.

The pair measurement in the supine and prone position corrects for geometrical and attenuation variations only approximately. Gamma-ray absorption is exponential and the geometrical sensitivity varies with the square of the distances between the source and the two crystals. Supine plus prone measurements apply a linear correction to these variations.

Further corrections for gamma-ray absorption and geometrical variations can be obtained from an analysis of appropriate phantom measurements. Figure 3 shows the spectrum of I^{131} gamma-rays for a point source placed 1, 4, and 7 inches below the surface of an 8 in. Maconite phantom. The counts in the photo peak (P) decrease exponentially. Adjacent to the photo peak, at lower energies, a region has been marked Compton (C). The count rate in this region does not decrease exponentially, since some of the primary gamma-rays which have been absorbed by small angle scattering appear as secondary gamma-rays in

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this region. It may therefore be assumed that the Compton region (C) can be used along with the photo peak (P) to derive the desired correction factor.

A Masonite phantom 8-in. thick, 12-in. wide and 5-ft. long was positioned on a horizontal plane such that its center line was 57 cm below the center of the crystals (Fig. 2). Spectral distributions of the count-rate for an ^{131}I source at various positions throughout the phantom were determined. The data were analyzed by adding the P and C values for points equidistant above and below the center line. The addition of these conjugate points produces data corresponding to the supine plus prone measurement of the source at any position within the phantom.

In addition, the photo peak (P) for the source at any position in the phantom was compared with the photo peak of the same source in air at the center position between the two crystals (Fig. 2). Then the ratio of count-rate in the photo peak area (P) from the source in the phantom to the count-rate of the photo peak area of the source in air, $P_{(\text{air})}$, is a measure of the effective gamma-ray absorption in the phantom.

In Fig. 4, the ratio P/C for each pair of measurements is plotted as a function of $P/P_{(\text{air})}$. The experimental points scatter around a mean curve (solid line), and most of them fall within one standard deviation from the mean (dotted lines). The latter corresponds to about a $\pm 5\%$ difference from the mean. Therefore, for any observed P/C value, the

absorption correction $P/P_{\text{(air)}}$ might have a 5% inaccuracy by using a mean correction curve.

Actually, the mean curve itself depends somewhat on the phantom thickness. The same measurements were repeated with a 5-in. and a 10-in. thick phantom and the mean curves for all three phantoms are shown in Fig. 5. The mean values for the 5-in. and 10-in. phantom fall within a few per cent from the mean values for the 8-in. phantom. Though some of the experimental points for the thinner and thicker phantoms fall outside the 5% range of the 8-in. phantom mean curve, it still appears to be a good approximation for the correction factor for all three phantoms (solid line in Fig. 5.).

Though the human body does not conform to any one of these phantoms, a combination of the three phantoms represent a fair approximation for various body weights and shapes. Only by giving patients of different body build known amounts of radioactivity, the adequacy of the average correction factor determined with the 8-in. phantom can be tested. Tables II and III show some examples of in vivo measurements. Applying the correction factor yields data which account for 100% of the dose at all times with an accuracy of $\pm 5\%$.

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FOCUSSING COLLIMATOR FOR 3-IN. DIAMETER CRYSTAL

For radioisotope distribution studies, scanners with 3-in. diameter sodium iodide crystals are commonly employed. The multi-hole focussing collimator attached to the crystal has its maximum sensitivity at about 2.5 in. below the front face of the collimator. This distance may be adequate for thyroid scanning, since the thyroid extends to only a few centimeters in depth inside the neck. Most organs, however, lie several inches below the body surface and extend frequently to or beyond the body midline.

By increasing the dimensions of crystal and focussing head, the weight of the required shielding becomes a major factor. However, a crystal mounted inside a steel room requires only additional shielding on the sides facing the subject. Therefore, it was decided to use one of the 8 x 4-in. sodium iodide crystals for body radioactivity distribution studies. Figure 1 shows one of the crystals inside the 3-in. thick lead shield which contains also the focussing head. The whole assembly can be raised and lowered and can also be moved to any position along the midline of the steel room. Twelve neg'ator springs* which counter balance the weight of the shield of about 500 pounds enable an effortless manual vertical movement of the whole shield-crystal assembly.

* Hunter Spring Company, Lansdale, Penn.

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For the scanning operation it becomes obvious that it is much simpler to move the patient back and forth and keep the focussing collimator stationary instead of the customary way of moving the collimator. Therefore a scanning bed was built* which allows selecting the scanning speed and the line width. The movement from side to side is so smooth that it is not objectionable to the patient.

Finally, the focussing collimator is constructed so that it can be lifted out of the lead shield after the crystal has been removed (Fig. 6). This allows one to have a set of collimators available which differ in the number of holes and the septa thickness between them. The collimator design has to be different depending on the gamma-ray energy and required resolution. In all cases the sensitivity should be as great as possible. Only the outside dimensions of the focussing collimator are fixed by the size of the hole in the lead shield. The height of the collimator, that is the distance between the front face of the crystal and the outside of the lead shield, is 5 in. (Fig. 7). The diameter of the collimator at the crystal face is 8 in. and at the opposite end it is 4 in.

As mentioned above, any one collimator cannot yield optimum results in all respects. Therefore, a collimator was designed for high sensitivity,

* Laboratory Associates, Inc., Belmont, Mass.

medium resolution for I^{131} gamma-rays and limited usefulness for high energy gamma-rays (Co^{60} , Fe^{59}). Figure 7 shows the dimensions of the 37 hole collimator chosen for this purpose. It was cast out of lead, though tungsten would be superior for high energy gamma-rays.

The isocount curves in air for a small I^{131} source are shown in Fig. 8. Only the counts from the photo peak area of the main gamma-ray line were registered. For comparison the isocount lines of a standard 19 hole collimator (Picker) for a 3-in. diameter crystal are also shown in Fig. 8. Its focussing point is at 2.5 in. from the collimator surface while the 37 hole collimator has the focal point at 4.5 in. The lines of 30, 50, 30 and 10 per cent sensitivity relative to that at the focal point penetrate to correspondingly deeper depths for the large collimator. Finally, the isocount lines obtained with the 37 hole collimator using only the outer ring of holes is also shown in Fig. 8. The depth penetration is slightly reduced and adjacent to the collimator, where the center sets of holes are closed, there is now an area of low sensitivity.

Though the present 37 hole collimator was not constructed for high energy gamma-rays, its isocount lines were determined for Co^{60} (Fig. 9). The volumes encompassed by the surfaces of different relative sensitivity are considerably enlarged compared with those for I^{131} (Fig. 8). A rather large fraction of the Co^{60} gamma-rays penetrates the collimator

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septas. If the collimator would be made of tungsten instead of lead, the difference between the I^{131} and Co^{60} isodose lines would be reduced.

Finally, the resolutions of the Picker 19 hole collimator for I^{131} and our 37 hole collimator for I^{131} and Co^{60} in the planes of their focal points are shown in Fig. 10. The full width at half of the maximum count-rate is 1.4 cm for the Picker collimator and 2.0 cm for the 37 hole collimator using I^{131} , while for Co^{60} it increases to 2.7 cm. This poorer resolution is further enhanced by the fact that even at rather large distances the count-rate does not fall below the 10 per cent level for the Co^{60} gamma-rays.

By placing I^{131} sources of different spherical volumes at the focal points of the 19 and 37 hole collimators, their absolute and relative sensitivity can be determined. Table IV gives the net counts per minute per microcurie of I^{131} for a 70-kev window at the 364-kev photo peak and the relative sensitivities of the two collimators. Using the 8-in. diameter crystals instead of the 3-in. one, about doubles the sensitivity. The small increase in sensitivity is due to the increase in collimator length and the increase in focal distance from 2.5 in. to 4.5 in.

Figure 11 shows an example of two kidney scans with Hg^{203} Neohydrin of the same patient with the 19 and 37 hole collimator. A 20 per cent cut-off level for background erase was used with the 19 hole collimator while no background erase is employed with the 37 hole collimator. Only

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from a larger group of patients scanned with the two collimators will it be possible to decide if the increase in focal depths results in any improvement of the scans.

SUMMARY

A whole body counter has been described which employs two 8 x 4-in. sodium iodide crystals inside a steel room. It allows the estimation of radioactivity in vivo independent of its distribution within the human body. This is accomplished by applying a geometrical and gamma-ray absorption correction to each measurement. The exact value of the correction factor is derived from an analysis of the measured energy distribution of the gamma rays incident on the two sodium iodide crystals.

One of the two crystals can be placed inside a 3-in. thick lead shield with a focussing head at the side of the crystal face. By placing a patient on a scanning bed, scans can be obtained with I^{131} comparable to those by moving a smaller crystal over stationary patients. The advantages of using the larger crystals are: increased focal distance, higher sensitivity and adaptability to high energy gamma-ray emitters.

Table I

Fraction of emitted gamma-rays of various energy that has interacted within tissue of given thickness.

Gamma-ray Energy (Mev)	Per cent absorption in tissue of thickness			
	5 cm	10 cm	20 cm	30 cm
0.279	46	70	92	95
0.364	44	68	90	97
0.511	37	61	85	91
0.662	35	57	82	92
1.25	27	47	72	85
1.46	26	45	70	83
2.0	22	40	55	70

* Since gamma-rays are absorbed in tissue mainly by Compton interaction, a scattered gamma-ray is produced each time. Even in cases where the scattered gamma-ray has very nearly the same energy and direction as the primary gamma-ray, the latter is regarded absorbed in tissue.

Table II

¹³¹I In Vivo Measurements
 (Subject's Weight: 165 lbs.)
 Height: 5'11"

Maximum I ¹³¹ Concentration	Time after oral dose (hrs.)	Uncorrected %	Corrected %
Stomach	0.0	45	100 [*]
	0.3	49	96
Blood and Thyroid	0.7	50	96
	1.0	51	96
	2.3	52	96
	70% excretion		
Thyroid	20.0	56	99

* Normalized after oral dose of iodide.

Table III

 I^{131} In Vivo Measurements

Maximum I^{131} Concentration	I. V. Doses	Subject I Wt. 108 lbs. Ht. 5'7"		Subject II Wt. 195 lbs. Ht. 5'11"	
		Uncorrected %	Corrected %	Uncorrected %	Corrected %
Blood	RISA	54	100*	46	100*
Blood	Rose Bengal	53	100	45	100
Blood & Liver	"	51	96	45	101
		52	98	44	101
Liver	"	50	96	44	101
Kidney	Hippuran	57	100	46	100
Diadur	"	50	95	43	103

* Normalized after I. V. dose of RISA.

Table IV

Comparison of collimator sensitivity for spherical I^{131} sources at the focal spot
(70-kev window at 364-kev line)

I^{131} -source (ml)	<u>Counts per minute-microcurie</u>		Relative sensitivity 37 and 19 hole collimators
	19 hole collimator 2.5 in. focal distance	37 hole collimator 4.5 in. focal distance	
0.25	4800	8300	1.7
1.0	4300	7800	1.8
10	2000	4600	2.3
25	1500	3900	2.6
50	900	2900	3.2

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Fig. 1. Two 8 x 4 in. sodium iodide crystals mounted on the backwall inside the iron room. Each crystal can be raised, lowered and tilted and moved along the center line of the iron room. One of the crystals has been placed inside a 3-in. thick lead shield with a multi-hole focussing collimator. The subject rests on a scanning bed.

Fig. 2. Schematic of two crystal whole-body counters as shown in Fig. 1. The numbers along a line corresponding to the center line of the patient represent the relative count-rate of a point source in air.

Fig. 3. Photo peak and adjacent Compton region of I^{131} pulse energy spectra. Data taken with I^{131} source under 1, 4, and 7 in. of Masonite. The count-rate in the photo peak decreases exponentially with increasing absorber thickness while the count-rate in the adjacent Compton region falls off much slower due to accumulation of small angle scattered radiation.

Fig. 4. ^{131}I source measurements in 8-in. thick Masonite phantom relative to source in air at the center position (Fig. 2). For each measurement the count-rates for the supine and prone position of the source in the phantom relative to the counters are added. Most values fall within one standard deviation from a mean curve.

Fig. 5. Mean curves of experimental data (Fig. 4) for I^{131} source in 5-, 8-, and 10-in. thick Masonite phantom.

Fig. 6. The 37 hole collimator has been removed from lower section of lead shield (Fig. 1). It can be replaced by a collimator of different design.

Fig. 7. Schematic of 37 hole lead collimator for 8-in. diameter crystal.

ISOCOUNT CONTOURS IN AXIAL PLANE
FOR I^{131} POINT SOURCE IN AIR

Fig. 8. Comparison of isocount curves for I^{131} in air with two types of collimators. On the left side, the 19 hole collimator (Picker) for a 3-in. diameter crystal with a focal depth of 2.5 in. In the center 37 hole collimator for 8-in. diameter crystal with a focal depth of 4.5 in. On the right side same 37 hole collimator but all holes closed by steel plugs except the 18 outer holes of the collimator.

ISOCOUNT CONTOURS IN AXIAL PLANE
FOR Co^{60} POINT SOURCE IN AIR

37HOLE COLLIMATOR FOR 8" DIAMETER CRYSTAL
THE 7 INNER HOLES CLOSED BY STEEL PLUGS

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Fig. 9. Isocount curves for Co^{60} source in air with 37 hole collimator.

COMPARISON OF 37 HOLE COLLIMATOR FOR 8" CRYSTAL
WITH 19 HOLE COLLIMATOR FOR 3" CRYSTAL

Fig. 10. Comparison of resolution of two collimators within their focal planes for I^{131} and Co^{60} sources (Figs. 9 and 10). The two inner curves are for an I^{131} point source with the 19 and 37 hole collimators; the outer curve is for Co^{60} with the 37 hole collimator.

Fig. 11. Kidney scan of a single patient with 19 and 37 hole collimators after a dose of 50 μ c of Hg²⁰³ Neohydrin. For the upper scan with the 19 hole collimator a 20% cut-off level was employed. No cut-off was used for the lower scan with the 37 hole collimator. Only a large series of scans could reveal differences in the performance of the two collimators.