Lung Counting: evaluation of uncertainties in lung burden estimation arising from a heterogeneous lung deposition using Monte Carlo code simulations.

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INTRODUCTION

The Human Monitoring Laboratory (HML), which operates the Canadian National Calibration Reference Center for *In Vivo* Monitoring (1), has replaced its phoswich detector lung counting system with a large area germanium detector system. The lung counting system consists of four 70 mm diameter x 30 mm thickness germanium detectors housed in a lead-lined steel room.

The HML has independently calibrated the detectors instead of routinely summing the data and treating them as a four-detector array. Other lung counting facilities generally calibrate their detectors as an array. Lung counters are calibrated by using a realistic torso phantom such as the Lawrence Livermore National Laboratory (LLNL) or Japanese Atomic Energy Research Institute (JAERI) phantom. The LLNL phantom (2) is based on a deceased LLNL plutonium worker and is larger than Reference Man (3). The JAERI phantom (4) is based on the smaller Japanese Reference Man (5). The counting efficiency of the detectors is determined by using lung sets that have the activity homogeneously distributed throughout the tissue substitute material.

A fundamental assumption made in calibrating a lung counter is that the deposition of radioactivity in the lung is homogeneous; however, depositions rarely follow this pattern. The distribution of the particles in the lung is a function of particle size, breathing rate (6), and health of the subject. Particles with a diameter of 1 μm will be partitioned between the bronchial region (upper), bronchiolar (lower), and alveolar interstitial (lowest) regions in a ratio of approximately 1:5:13 in a healthy subject at rest; similarly, particles with a diameter of 5 μm will be partitioned in a ratio of approximately 1:1:1.5. These ratios will change depending on the particle size and the level of activity of the subject.

There will be an uncertainty on the estimate of activity in the lung if the lung counter is calibrated using a lung set that has the activity homogeneously distributed throughout the tissue substitute material and the deposition in the subject is heterogeneous. Preliminary (7) data obtained in the HML suggested that individual detector calibration minimized this uncertainty; however, this study was performed with a limited number of source distributions and at only two energies: 63 and 93 keV. This paper has extended that study using Monte Carlo code simulations to model the HML's detectors with a phantom derived from the MIRD (Medical Internal Radiation Dose) and LLNL phantoms.

MONTE CARLO METHODS

The Monte Carlo method is a mathematical technique for solving a problem that is dependent upon probability in some manner. The technique is useful when exact formulation describing a process may be too difficult, or even impossible, to derive and solve by direct methods.

Monte Carlo N-Particle (MCNP) transport code system is a general purpose radiation transport code that is the result of 50 years of developmental efforts. The release version used in the calculations described below is MCNP4.2A. This code is robust and over the period 1991-1993 it was employed by thousands of users. During that time no important errors were found in the code. The code has an extensive library of continuous-energy cross-section data. Photon cross-section tables span the energy range 1 keV - 100 MeV, and were extracted from the Evaluated Nuclear Data File.

MCNP can give the normalized number of particles interacting with a scintillation detector, which essentially the counting efficiency of the detector measured in units of counts per photons. This is equivalent to an experimentally determined "absolute peak efficiency" of the detector. The tally is also accompanied by a relative error that is approximately equivalent to the inverse square root of the number of histories contributing to the tally. Therefore, the precision is improved at a rate equal to the square root of the increase in histories sampled. A two-fold improvement in the precision would require sampling four times as many histories. The authors of MCNP consider that a relative error value of 0.1 - 0.2 suggests that the tally result is questionable. Tally results for which the relative error is above 0.2 are not likely to be meaningful, but are generally reliable for a relative error less than 0.1. Except where noted, all the simulations described below have had relative errors less than 0.1.

THE SIMULATIONS

The germanium detectors were modeled based on specifications supplied by EG&G Ortec. They resemble the germanium detectors currently used in the HML. Detector #1 was placed over the upper part of the left lung; detector #2 was placed over the lower part of the left lung and the heart; detector #3 was placed over the upper part of the right lung; detector #4 was placed over the lower part of the right lung.

The virtual torso was modeled on the torso section of the MIRD phantom, but the lungs were modified to be more consistent with the size and shape of the lungs in the LLNL phantom. Each virtual lung was divided into 8 sections, giving 16 sections in all, to approximately correspond to a lung set manufactured by Pacific Northwest National Laboratory for the HML. Each lung was divided in the middle in each of the x-y, x-z, and y-z planes, the origin being set to the center of each lung. The lungs were filled with lung tissue, the organs were filled with muscle tissue, the ribs were filled with skeletal bone, and the torso cavity was filled with soft tissue. The compositions of these tissue were taken from ICRP 44.

Activity was placed into sections of the lungs in various arrangements. The reference position is *Lung 0* where the activity is homogeneously distributed throughout the lung. For each lung configuration, photons of the following energies were generated and followed: 17 keV, 20 keV, 40 keV, 60 keV, 120 keV, 240 keV, 660 keV and 1000 keV. The photons that interacted with the germanium in the detectors were tallied individually so that an individual detector efficiency was obtained for each configuration. An array was simulated by simply adding up the individual detector tallies for a given configuration.

Each run was performed on a single energy using 10,000,000 photons. The relative error was less than 0.1 for all simulations. The 17 keV simulations were run using either 50,000,000 or 100,000,000 photons to meet the relative error requirement; in some cases this requirement was not met and the efficiency was set to 0.0. The 17 keV simulations are, therefore, discussed separately.

RESULTS AND DISCUSSION

The data obtained from the MCNP simulations was analyzed in several ways: 1) comparison of the counting efficiency obtained for a heterogeneous deposition to a homogeneous deposition; 2) comparison of the array to individual detectors; 3) evaluation of the contribution of different sections of the lungs to the detector response was evaluated.

Counting Efficiency.

The efficiency data for a four detector array is discussed below. Individual detector responses will be discussed during the presentation.

Array (Energy vs Efficiency) 17 keV: The array efficiency has been calculated by adding the efficiencies of the four detectors for each lung configuration. In some cases the MCNP result did not satisfy the relative energy requirement, but these counting efficiencies are all very low. They have been set to zero for the purpose of obtaining a simulated array efficiency.

Some configurations had a zero counting efficiency. The activity in these configurations is in the back sections of the lung. *Lung 14,* with activity in the bottom-front-right of the right lung, has the highest counting efficiency.

Array (Energy vs Efficiency) 20 keV - 1000 keV: An array of four detectors was simulated by adding the efficiencies of all each detector for each lung configuration. This results in a summation and averaging. For example, the array counting efficiency for *Lung 0* was 0.0087 cnt/photon at 60 keV. The same counting efficiency for the individual detectors (1 to 4) at 60 keV was 0.00206 cnt/photon, 0.00187 cnt/photon, 0.00247 cnt/photon, and 0.00233 cnt/photon respectively.

The effects of the other configurations are moderated by the array, and the differences in counting efficiency compared to *Lung 0* are less than for the individual detectors. At each energy there is a maximum value for the counting efficiency (when the activity is concentrated close to the detector) and a minimum value (when the activity is remote from the detector). These values can be expressed as a factor of the *Lung 0* counting efficiency.

The table below shows the "maximum factor" obtained by dividing the maximum efficiency at each energy and for each detector assembly (individual and array) by the *Lung 0* efficiency at the corresponding energy. These values correspond to the amount by which the activity could be underestimated.

The table also shows the "minimum factor" which is obtained by dividing the *Lung 0* efficiency at each energy and for each detector assembly (individual and array) by the minimum efficiency at the corresponding energy. These values correspond the amount by which the activity could be overestimated. Some columns have "N/A" listed which means that no counts were detected. In other words the deposition will be missed and the activity will be assigned a zero value, which would be a gross underestimate.

The factors for the array assembly are always less than for an individual detector, except in the case of 17 keV photons when "N/A" is the result, showing that the previous finding (7) was, in general, not correct.

Contribution of different lung sections to detector response.

The relative contribution of each of the sixteen lung sections has been evaluated by comparing the number of photons originating from a given lung section, and depositing the full amount of energy in a given detector, to the total number of photons originating in all lung sections, and depositing the full amount of energy in the same detector.

It is interesting to note that the contribution from activity in a lung opposite to a detector (i.e., activity in left lung contributing counts to the upper right or lower right detectors) only rises to 26% at the highest energy, 1,000 keV. At 20 keV and lower there is essentially no contribution; the contribution rises to approximately 10% at 40 keV and 15% at 60 keV.

Similarly, one can see that the contribution from the other section of the lung over which a detector is placed (i.e. contribution from the upper half of the lung to the detector placed over the lower half) rises from 15% at 17 keV to 30% at 1,000 keV. This demonstrates that the detectors view for low energy photons is mostly comprised of the lung material directly below or in close proximity, i.e, within a few centimeters. At higher energies the view expands somewhat, but the main influence is still the material directly below the detector. This is further exemplified if one looks at the contribution of activity from the back half of the lung to the detector that is placed directly overhead. This contribution rises from approximately 1% at 17 keV to 13% at 1,000 keV.

At 17 keV the TFL, BFL, TFR and BFR sections contribute 79% of the total counts, clearly demonstrating the geometry effects of low energy photon counting. In many instances the back sections contribute less than 1% of the counts to the array's total. The contribution of the TFL, BFL, TFR and BFR sections drops to 41% at 1,000 keV with the other sections contributing between 3 - 8% of the total counts. *The lung sections directly below the detectors have a large influence on the derived activity.*

CONCLUSIONS

This study has shown how the deposition pattern of an aerosol inhalation can affect the results obtained by measuring the activity with a four-detector array. It has also resolved the question of whether detectors should be individually calibrated or treated as an array. Detector arrays will minimize the uncertainties arising from the geometry of the lung deposition and should be the preferred counting configuration.

Low energy photon counting can give rise to large uncertainties. The uncertainty on the activity estimate of an internal deposition that emits 17 keV photons can be underestimated by a factor of four, overestimated by a factor of forty, or missed completely. As the photon energy rises to 60 keV the uncertainty on the activity decreases so that the maximum overestimate (or underestimate) will be a factor of three. As the energy rises to 1,000 keV this factor drops to 1.8.

The results of the simulations have shown that the detectors view only the lung sections in close proximity at low photon energies. As the photon energy rises the "view" expands, but the lung sections that are in close proximity to the detectors remain the dominant contributors to the observed counts. This shows that calibrations based on homogeneous depositions are, at best, a poor approximation of reality. However, this calibration is repeatable and allows facilities to either intercompare their results in a consistent manner with other facilities, or participate in intercomparisons designed to validate their activity estimate protocol.

REFERENCES

1. Kramer, G.H. and Limson Zamora, M. *The Canadian National Calibration Reference Centre for Bioassay and In-Vivo Monitoring: A Program Summary.* Health Physics 67(2), 192-196 (1994).

2. Griffith R.V., Dean P.N., Anderson A.L., Fisher J.C. *Tissue equivalent torso phantom for intercalibration of in vivo transuranic nuclide counting facilities.* In: Advances in Radiation

Monitoring, Proceedings of an International Atomic Energy Agency Conference. IAEA-SM-229/56, 4493-504. IAEA, Vienna (1978).

3. International Commission on Radiological Protection. *Report of the Task Group on Reference Man.* Oxford: Pergammon Press; ICRP Publication No. 23 (1975.).

4. Shirotani T. *Realistic torso phantom for calibration on in vivo transuranic-nuclide counting facilities.* J. Nucl. Sci. and Tech. 25(11), 875-883 (1988).

5. Tanaka G., Kawamura H., Nakahara Y. *Reference Japanese man I - mass of organs and other characteristics of normal Japanese.* Health Physics. 36, 333-346 (1979).

6. Annals of the International Commission on Radiological Protection. *Human respiratory tract model for radiological protection.* Oxford: Pergammon Press; ICRP publication No. 66 (1993).

7. Kramer G.H., Burns L.C. *Effect of radionuclide distributions on lung counting efficiency.* Rad Prot. Dosim. 61(1-3) 145-147 (1995).

Table: Factors by which the efficiency differs from that obtained from the *Lung 0* configuration. N/A means that the lowest counting efficiency was zero.

	Highest efficiency divided by *Lung 0* efficiency							
	17 keV	20 keV	40 keV	60 keV	120 keV	240 keV	660 keV	1000 keV
Detector 1	11.2	10.0	5.1	4.3	4.4	4.1	3.5	3.2
Detector 2	11.6	8.4	5.8	5.2	4.9	4.3	3.5	3.2
Detector 3	8.6	8.5	4.8	4.6	4.4	4.1	3.6	3.5
Detector 4	10.5	7.8	5.6	5.0	4.8	4.6	3.9	3.5
Array	3.8	3.3	2.3	2.2	2.1	2.0	1.8	1.8

	Lung 0 efficiency divided by lowest efficiency							
	17 keV	20 keV	40 keV	60 keV	120 keV	240 keV	660 keV	1000 keV
Detector 1	N/A	N/A	9.7	7.2	6.8	4.8	3.8	4.0
Detector 2	N/A	N/A	10.1	7.4	6.7	5.8	4.4	3.3
Detector 3	N/A	N/A	15.7	9.3	9.3	7.5	5.3	4.7
Detector 4	N/A	N/A	13.2	9.6	8.5	7.2	4.1	3.5
Array	N/A	26.0	3.3	3.0	2.7	2.7	2.4	1.8