MCNP evaluation of the effect of chest wall thickness, tissue composition, and photon energy on the quantity Muscle-Equivalent-Chest-Wall-Thickness.

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INTRODUCTION

The measurement of radioactivity deposited in the lung is often performed with an array of germanium detectors placed over the chest of the individual. The detector array size can be from four to eight detectors depending on the size of the germanium detector and Dewar assembly. Typically, large area diameter germanium detectors (70 mm diameter) consist of four detectors, and the larger arrays consist of smaller diameter detectors (50 mm diameter).

These arrays 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 is based on a deceased plutonium worker who worked at LLNL and is larger than Reference Man. The counting efficiency of the detectors is usually determined by using lung sets that have the activity homogeneously distributed throughout the tissue substitute material.

As photons pass through the chest wall they are attenuated to differing degrees (based on their energy) by the tissues present: muscle, adipose, cartilage, bone (rib). For example, 1 cm of bone will allow no transmission of 17 keV photons, 57% of 60 keV photons, and 89% of 1MeV photons; by contrast, 1 cm of adipose tissue will allow the transmission of 47% of 17 keV photons, 83% of 60 keV photons, and 93% of 1MeV photons. As a result, lung counters must be calibrated in terms of energy, chest wall thickness (CWT), and at low energy by the tissue composition of the chest wall. Both the LLNL and JAERI phantoms have accessories that allow a limited simulation to be performed. The limitations are that both phantoms have a fixed amount of rib and cartilage. The CWT and adipose-muscle content of the chest wall can be varied by the use of overlay plates that are of different thicknesses and are made of different tissue substitute materials; however, the range of thicknesses and adipose-muscle ratios is limited. The variable, adipose mass fraction (AMF) can be eliminated by the use of a derived quantity: Muscle-Equivalent-Chest-Wall-Thickness (MEQ-CWT) and thus simplify the relationship.

Monte Carlo simulations have the advantage that experiments can be performed when either the experimental equipment is unavailable, or does not exist. For example, neither torso phantom allows the bone equivalent material to be removed from or altered in the phantom's chest plate cover. This study has simulated the Human Monitoring Laboratory's lung counting system that consists of four 70 mm diameter x 30 mm thickness germanium detectors and a torso phantom with differing CWT and AMF values. The effect of rib on the MEQ-CWT has also been assessed.

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. The Monte Carlo method constructs a stochastic model representing the process of interest. A set of high quality random numbers is then used to sample the probability distribution functions defined by the model. The result is an estimate of a physical quantity characteristic of the process, specified with a measured degree of confidence.

THE SIMULATIONS

The germanium detectors were modelled based on data supplied by EG&G Ortec. They resemble the detectors currently used in the HML. The torso was modelled on the LLNL torso phantom. The organs were filled with muscle tissue, the lungs were filled with lung 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 ICRU 44. The lungs were based on the dimensions of the LLNL lung sets held in the HML. The CWT was altered from 1.63 to higher values by expanding the dimensions of the outer torso envelope. In this way, CWT values of 1.63, 2.0, 2.5, 3.0, 3.5, 4.0, 4.5, 5.0, and 5.5 cm were generated. The tissue composition was altered by changing the isotopic composition and density. Activity was loaded into the lungs to simulate a homogeneous distribution of radioactivity and photons of the following energies were generated and followed: 17 keV, 40 keV, 60 keV, 120 keV, 240 keV, 660 keV and 1000 keV.

MUSCLE EQUIVALENT CHEST WALL THICKNESS

The MEQ-CWT is the thickness of muscle-equivalent-absorber that reduces the photon flux from the lungs by the same amount as the actual combination of muscle and adipose tissue in the chest plate and overlay plates. This method has been successfully used elsewhere for the removal of the adipose

$$MEQ - CWT = \frac{CWT}{m_{msc}} \left[\frac{m_{adp}\, AMF}{100} + \frac{m_{msc}\,(\,1 - AMF\,)}{100} \right]$$

mass fraction as a variable. The function is shown below:
where MEQ-CWT is the muscle equivalent chest wall thickness (cm), CWT is the physical chest wall thickness (cm), AMF is the adipose mass fraction (%) or adipose-muscle ratio expressed as a percentage, μ_{adp} is the linear attenuation coefficient at a given energy for adipose tissue (cm^{-1}), and μ_{msc} is the linear attenuation coefficient at a given energy for muscle tissue (cm^{-1}).

Equation 1 shows that MEQ-CWT changes as the photon energy increases from 17 keV to 1000 keV. This is because the linear attenuation coefficients change with photon energy. Both μ_{adp} and μ_{msc} are obtained from the literature by multiplying the mass attenuation coefficients by the appropriate density (adipose or muscle). Both μ_{adp} and μ_{msc} vary greatly with energy and in order to calculate them at any energy they have been fitted to the following function:

Where: μ is the linear attenuation coefficient corresponding to the amount of adipose tissue and muscle tissue present, and x is the photon energy (keV).

RESULTS AND DISCUSSION

At 17 keV the tissue composition of the chest wall can greatly affect the transmission of the photons. Taking a realistic range of AMF values to be 10% to 50% one can see that the counting efficiency at the smallest CWT (1.63 cm) can change by a factor of approximately 1.5. The change in counting efficiency increases as the CWT becomes larger so that at a large value of CWT, say 4.0 cm, the change in the counting efficiency is approximately a factor of 2.2. As the photon energy rises there is much less dependence on the tissue composition of the chest wall. For example, the counting efficiency at the smallest CWT changes by a factor of approximately 1.04. As for 17 keV photons, the change in counting efficiency increases as the CWT becomes larger, so that at a CWT of 4.0 cm the change in the counting efficiency is approximately a factor of 1.1.

These difference diminish as the photon energy rises. Table 1 summarises the change in counting efficiency, as a function of photon energy, for CWT values of 1.63 cm and 4.0 cm as the tissue composition of the chest wall changes from 10% adipose to 50% adipose. It can be seen that at 60 keV and above, the composition of the chest wall can be neglected as a factor, but that at lower photon energies it must be considered as a variable that could contribute a significant uncertainty to the counting efficiency.

Table 1: Factor - change in counting efficiency when AMF changes from 10% to 50% for two CWT values and as a function of photon energy

	Energy (keV)						
CWT	17	40	60	120	240	660	1000
1.63	1.47	1.04	1.02	1.02	1.02	1.01	1.02
4	2.22	1.10	1.05	1.04	1.02	1.01	1.01

The derived quantity, MEQ-CWT, eliminates the parameter AMF as described above. Data plotted for counting efficiency as a function of MEQ-CWT transforms the family of curves at a given energy to a single curve which has an exponential dependence on the MEQ-CWT. This is particularly true for the 17 keV data.

Although the use of MEQ-CWT eliminates the dependency of AMF on the counting efficiency, two limitations still appear to remain. First, the AMF in the subject must be accurately assessed otherwise an incorrect counting efficiency will be used. Second, and perhaps more importantly, the effect on the counting efficiency due to the difference of the subject from the calibration phantom for: the rib spacing, size of ribs, amount of bone to cartilage, and bone density - this is termed the *bone factor*.

An estimate of the *bone factor* has been made by replacing all the bone material in the virtual phantom by tissue of the same composition as the rest of the chest wall. In other words, the bones have been removed and replaced with the appropriate tissue. While this is recognised as an unreasonable and unrealistic scenario, it does provide some insight into the magnitude of the *bone factor*. This has been done for 17 keV, 40 keV and 60 keV photons. The higher two energies show that the absence or presence of ribs in the chest wall has essentially no influence on the counting efficiencies derived as a function of MEQ-CWT. Typically, the counting efficiency at a given MEQ-CWT is a factor of 1.12 higher when the ribs are removed at 40 keV and a factor of 1.04 at 60 keV. This shows that small differences in the bone content of the chest wall of a subject compared to the calibration phantom will have no influence on the derived counting efficiency. However, at 17 keV the difference between the counting efficiency for a chest wall with rib and no-rib is a factor of two. This implies that small differences in the bone content of a subject compared to that of the calibration phantom will have a small influence on the counting efficiency. This difference is likely to be no more than 20%.

CONCLUSIONS

Monte Carlo simulations have been used to derive the relationship between the counting efficiency and both the CWT and AMF as a function of photon energy. AMF can be eliminated by the use of the derived quantity MEQ-CWT and thus simplify the relationship. As a result, the counting efficiency at any AMF can be predicated from a single exponential plot of counting efficiency against MEQ-CWT; therefore, it is unnecessary to have a family of curves for different AMF values. This methodology eliminates the need for interpolating a counting efficiency for a subject whose AMF does not fall directly on a calibration curve.

This work has also shown that it is unnecessary to build torso phantoms that have a large variety of overlay plates of differing tissue composition, which at best can only simulate a limited number of values. The use of MEQ-CWT shows that only one calibration curve using 0% AMF is required. This curve is then sufficient to define a detector calibration as a function of CWT at any value of AMF that a subject may possess simply by converting the subject's CWT to MEQ-CWT. Phantom production should, therefore, be simplified as less materials would be required for the manufacture. Overlay plates of differing AMF values would no longer be required, although overlay plates of different thickness are still needed to define the calibration curve.

Uncertainties due to the measurement of the subject's AMF and the difference between the subject's chest (ribs, cartilage) compared to the phantom cannot be eliminated by the use of MEQ-CWT. These factors will contribute to the overall uncertainty in an activity estimate made from a lung count. However, subject to phantom difference is likely to be small with respect to bone content of the chest wall. AMF measurements are only important for 17 keV photons and can be neglected at photon energies greater than 40 keV.