The term "dosimetry" should be replaced with the phrase "dose estimation" in the case of internal emitters. Direct measurement of absorbed dose (D) within an organ or tumor, via thermoluminescent (TLD) crystals or other solid-state devices, is problematic in a clinical context. Associated trauma alone is sufficient to invalidate any resultant readings. Also the possibility of tumor cell migration makes such attempts questionable. We note that similar conclusions hold for brachytherapy and external beam radiation. Two types of dose estimation are seen in nuclear medical literature: those applied to standard geometric phantoms (Type I calculations) and those applied to specific patients (Type II). In the former case, the estimates are conventionally done for one of two reasons. These involve either applications to regulatory agencies or a comparison of several radiopharmaceuticals that have been designed to target the same organ, cell type or molecule. Type I computations are, by far, the most commonly done by medical physicists. In the calculations, it is important to realize that D is a vector quantity that gives the mean organ dose. An associated vector ($\tilde{A}$) gives the time-integrated activity in a set of source organs. Typically, D is the dimensionally larger vector clinically since only a few source organs can be identified in a set of patient images. In the traditional Type I calculation, the rectangular matrix **S** connects the two vectors by the equality: $D = \mathbf{S} * \tilde{A}$. Here, **S** has been previously determined using Monte Carlo (MC) methods and one or another standard phantom geometry. Up to 10 phantoms are available in the MIRDOSE3 program of Stabin. Use of radiopharmaceuticals for patient (pt) cancer therapy has led to increasing interest in Type II estimates. Either beta or alpha rays (non-penetrating radiation) will be the dominant terms in such treatments so that **S** is diagonal if we neglect the photon contributions. In this context, we can modify a standard **S** matrix for the corresponding phantom by multiplying by a ratio of organ mass (m) values for the target (source) organs. Since m is in the denominator of **S**, the correction is: $\mathbf{S}(pt) = \mathbf{S}$ (phantom)*m(phantom)/m(pt). These corrections can be very large – up to factors of two-fold or more and are provided by the pt CT or MRI images. Type II computations may eventually be made by MC methods directly using the anatomic imaging results. Point source functions for the radionuclide of interest can also be used. As in the case of brachytherapy or external beam, spatial distributions of dose would be an additional result of such voxel-based estimates. In this case, benefit and risk would be evaluable using pt-specific dose-volume histograms.

Educational Objectives:

1. Realizing that radionuclide dose is an estimated not a measured quantity.
2. Understanding that two types of internal emitter dose estimates are made; one for phantoms and another for specific patients.
3. Knowing that the standard formula for dose (D) is $D = S*\tilde{A}$.