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Monte Carlo Basics for Radiation Dose Assessment in Diagnostic Radiology

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Introduction

Monte Carlo simulations are the basis of all modern x-ray dosimetry methods in diagnostic radiology. Monte Carlo (MC) methods are different from the larger class of computer simulation techniques in that they explicitly compute stochastic events and track the outcome. For x-ray dosimetry, MC methods track the trajectory of individual x-ray photons, one by one, as they exit an x-ray source, enter the simulated patient anatomy, and undergo scattering and absorption events. The amount of energy deposited in the “patient” is tallied at each location where a dose-deposition interaction takes place. Typically, millions to billions of photon histories are computed and the energy deposited in each volume element (voxel) in the mathematical phantom is then divided by the tissue mass of that voxel, resulting in the absorbed dose for the voxel – defined as imparted energy per unit mass.

Modern computers are very fast and billions of photon histories can be realistically simulated to estimate the radiation dose deposited to anatomy from a given radiological imaging application. Despite this large number of simulated photons (e.g. 10^9), actual x-ray imaging involves 10^{14} to 10^{16} photons for each mammographic or CT acquisition, respectively – a factor of 10,000 or more greater than what is possible in most MC experiments. Thus, it is common to also record the air kerma at the entrance of the phantom for radiographic or mammographic applications, or the air kerma at the center of the field for

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computed tomography applications. In this way, a coefficient representing the absorbed dose per unit air kerma – in the interesting units of mGy/mGy, is computed. In the old days of radiology, these coefficients used different units and were sometimes called the “roentgen to rad conversion factors”. These coefficients allow dose levels in actual imaging procedures to be estimated using physically-measured air kerma levels in the radiography room or CT suite.

Monte Carlo Methods in General

One of the key components of MC software is the random number generator (RNG). Just like the flip of a balanced coin returns integers of either 0 or 1 (heads or tails), the RNG is a computer subroutine which typically returns pseudo-random real numbers between 0.000 and 1.000, with a completely uniform distribution. For MC computations involving x-ray attenuation and scattering, other more complicated statistical distributions need to be used. In this case, the raw RNG can be converted to different statistical distributions analytically or by sampling techniques (*e.g.* the “rejection method”). For x-ray propagation, the distributions of interest include exponential attenuation, the relative probabilities of different interaction types, and the tissue-type and x-ray energy-dependent scatter angle distributions specific to Rayleigh and Compton scatter.

The x-ray spectra used in clinical radiological imaging are poly-energetic. Therefore, for MC dose calculations to be accurate, they need to consider the x-ray spectrum used for the specific application under investigation. In this case, tabulated x-ray spectra are used (1,2). In general, a MC computer run is performed for each mono-energetic x-ray beam energy independently, using energies ranging from the minimum photon energy to the maximum photon energy in the x-ray spectrum. For CT applications, for example, this means MC simulations run from about 15 keV to 140 keV, by 1-keV intervals. The clinically relevant poly-energetic dosimetry data is spectrally weighted from the raw mono-energetic dose data.

MC Dosimetry for Specific Modalities

Breast Imaging

MC techniques have been used extensively for determining dose coefficients in mammography, which are called the *normalized glandular dose* coefficients (DgN). The dosimetry considerations in mammography are unique because only the dose to the glandular tissue in the breast is computed – dose to the skin and adipose tissue in the breast (the majority of the volume of the breast) is not included in the dose metric because they are not the tissues at risk for breast cancer. Traditionally, a simple “D” shaped compressed breast is modeled, and other than a layer of skin, the entire breast is considered to be a homogeneous mixture (*i.e.* perfect mixing) of adipose and glandular tissue. In the past, the average breast was assumed to be 50% adipose and 50% glandular tissue, with breast densities ranging from 0% to 100%. More recent analysis of the breast from 3-dimensional breast CT and other data have shown that the median breast density is about 17% glandular by volume (3), and these data sets have also demonstrated that glandular tissue is not homogeneously distributed but is more centrally located (4). Using this more accurate

heterogeneous model of breast density, recent MC studies have shown that the radiation dose in mammography is about 30% lower than previously assumed (5).

Computed Tomography

MC dosimetry in computed tomography (CT) benefits from the fact that CT images are essentially three-dimensional maps of the (rescaled) linear attenuation coefficients of the tissues in the patient's body. This means that clinical CT image data sets can be used as accurate mathematical models in MC simulations. While CT dosimetry can be used to compute absorbed dose to all tissues in the scan, often the dose to specific organs is computed (6, 7) – this requires the CT images to be segmented, *i.e.* the boundary of each organ needs to be outlined either manually or algorithmically. Individual organ dose data is necessary for the estimation of *effective dose*, where the absorbed dose for each organ is multiplied by an organ-specific tissue weighting factor. A sum of the weighted organ dose values over all exposed regions leads to a risk estimate (the effective dose) for that CT procedure. Modern CT systems make widespread use of tube current modulation, which changes x-ray tube output continuously throughout the CT scan; making CT dosimetry more complex. However, adjustment for this in software-based dosimetry systems can restore accuracy (8).

Radiography

MC computed dose coefficients for radiography were computed in the U.S. and Europe several decades ago, but unfortunately there are few new studies in this area. Because radiographic techniques (*e.g.* tube potential, beam filtration, *etc.*) have changed over the years with the advent of digital imaging, these data need to be updated. Because CT image data is so widely available and would be useful for simulating anatomy for radiographic projections, there is a pressing need for MC scientists to engage in a comprehensive reassessment of dose tables for the dozens of radiographic projections that are used clinically. Both absorbed dose and (organ-dose based) effective dose studies are sorely needed for the wide array of radiographic projections used in modern radiology clinic.

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Summary

Monte Carlo studies are the basis of all modern x-ray dosimetry, because these techniques can and do provide accurate and comprehensive dose tables (typically as coefficients expressing *absorbed dose per unit air kerma*). The speed of modern desktop computers has allowed widespread assessment of dose coefficients across a range of clinical x-ray imaging procedures, however more work is required to update dose tables for many applications, especially radiographic imaging studies.