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Using simulations of the detector performance for enhanced image reconstruction in molecular imaging

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ABSTRACT

In Emission Tomography images are reconstructed by solving an inverse problem: the three-dimensional activity map producing the signal observed by the gamma camera or the Positron Emission Tomography (PET) detector is estimated given a model of the imaging system response. This model gives the set of probabilities $\mathbf{R}_{ij}$ that a γ or $\beta+$ emission occurring at point j in the volume of interest be detected in detection element i. Thorough modeling of this **R** system matrix (SM) is essential for ensuring the most accurate possible estimate of the activity distribution within the object of interest. Thirty years from now, it was proposed to calculate the system matrix **R** based on Monte Carlo simulations, as opposed to using analytical geometrical models, for increased accuracy. A lot of progress has been made since the initial idea and using simulations for enhanced SPECT and PET image reconstruction has become a reality. In this paper, we review the rationale for this approach, explain the advantages and limitations, the performance that can be achieved, and the challenges that remain to be solved.

1. Introduction

In Single Photon Emission Computed Tomography (SPECT) and Positron Emission Tomography (PET) molecular imaging, images are obtained by solving an inverse problem, that is by estimating the unknown γ or $\beta+$ radiotracer activity distribution that is best compatible with the measured data, given a model of the imaging system response function. In a discrete formalism, this corresponds to solving the following equation:

$$p = \mathbf{R}f \quad (1)$$

where p represents the projections or the sinograms arranged as a 1D vector, f is the activity distribution to be reconstructed also arranged as a 1D vector and **R** is a 2D matrix, called the system matrix (SM), that describes how f, in the so-called “image space”, is transformed into p in the projection space. Each entry $\mathbf{R}_{ij}$ represents the probability that a γ photon or a $\beta+$ particle emitted in voxel j of the object be detected in projection bin i. The quality of the reconstructed images is directly affected by the ability of the SM to accurately describe the response of the imaging system. For a long time, the entries of this SM have been calculated using a line integral model accounting for a simplified description of the detection geometry [1,2]. Yet, this is an idealized model for emission tomography (ET), both from the geometric and the physics points of view. Indeed, the line integral model assumes that the observed flux of photons arriving in one bin of the detector is only due to activity along an infinitesimally narrow line, which is obviously a crude assumption in SPECT since this would assume a perfect collimation (no divergence, no spatial spread, no collimator penetration). This is approximate in PET as well since the line of response (LOR) defined by two crystals is actually a tube of response including many possible narrow LOR. Some more sophisticated geometrical analytical models have been proposed [3,4]. The simplified line integral or more sophisticated geometrical approximations also neglect some important physics aspects that are inherent to ET, including scatter and attenuation in the object under investigation, interactions within the collimator in SPECT, positron range in PET, or particle interactions within the detector crystals. For the reconstructed f activity distribution to accurately estimate the actual activity distribution, the SM should precisely reflect the real probability controlling the physical experiment, ideally including all physics and geometrical effects.

As soon as 1985, Floyd et al [5,6] suggested that a more realistic model of a SPECT acquisition be used for SM calculation and proposed to have that model established using Monte Carlo simulations. Indeed, at that time, Monte Carlo (MC) simulations were

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already shown to accurately reproduce SPECT acquisitions and offered the possibility to account for many physical characteristics of the acquisition system (energy window setting, energy and spatial resolutions, radius of rotation defining the solid angles through which a given point in the object is seen by the detector) and for the object-dependent scattering medium in which the photons propagate and undergo scatter and attenuation. The implementation as proposed by Floyd et al was only in 2D, ie reconstruction of a 2D image including 1,026 voxels, from a tomographic acquisition of 180 views of 64 measurements, for computation time reasons. The same idea of MC-based SM calculation was proposed for PET in 1988 [7] with a complete modeling of Compton scattering, detection efficiency, attenuation, positron range and non-collinearity of the annihilation photons, still in 2D. The extension to fully 3D reconstruction was introduced in 2004 for Tc-99m SPECT by Lazaro et al. [8,9], with MC calculation of the SM modeling scatter and attenuation in the object and accounting for the detector response function (DRF). This initial report showed a definite advantage of iterative reconstruction using the MC-based SM compared to iterative reconstruction accounting for attenuation, scatter and DRF using an analytical SM model (Fig. 1). In PET, the use of a MC-calculated SM in fully 3D was reported first for small animal imaging [10], modeling the geometric response of the system and photon scattering within the detector. Based on these seminal studies that demonstrated the feasibility of accurate SM calculation based on comprehensive 3D MC modeling of object and/or detector features, many investigators have extended the original ideas, evaluating the benefit of making the SM more accurate, and proposing solutions to overcome the hurdles associated with the practical implementation of this approach.

To give an overview of the importance of simulations for accurate image reconstruction in PET and SPECT, the outline of the paper will be as follows. Section 2 will explain how the simulation of the detector performance and of the whole imaging settings can offer an elegant approach for quantitative image reconstruction in ET. Section 3 will present the challenges associated with this approach, while Section 4 will discuss the applications investigated so far and associated performance, before drawing some conclusions.

2. How can simulations contribute to accurate image reconstruction?

2.1. Deriving SM using simulations

SM calculation requires each $\mathbf{R}_{ij}$ entry of the SM to be estimated. This probability that a γ photon or a positron emitted in voxel j of the object be detected in projection bin i depends not only on the type of particle, and on the detector features, but also on the object properties, which impact the interactions that

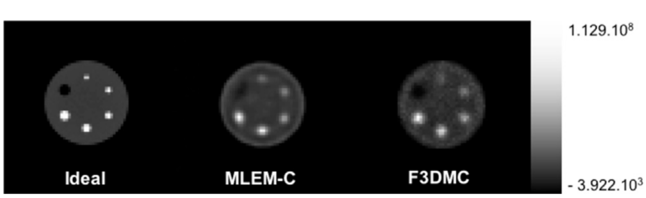

Fig. 1. Very first study demonstrating the feasibility of 3D SPECT reconstruction using a system matrix (SM) calculated using Monte Carlo (MC) simulations. Qualitative comparison between a simulated SPECT image reconstructed using MLEM involving either an analytical SM including attenuation and point spread function correction applied on scatter corrected projections (MLEM-C) or using a SM estimated using MC simulation (F3DMC). The cylindrical inserts are less distorted using the Monte Carlo SM. Adapted from ref [9] in which all reconstruction details can be found.

particles emitted in the object will undergo. Therefore, for a given acquisition protocol involving a specific radiotracer and detector, a different **R** should ideally be calculated for each patient (or animal). The major advantage of using a simulation approach to calculate **R** is that every phenomenon involved in the image formation process can a priori be accounted for in the reconstruction as long as it can be modeled using simulations. The simulation-based SM calculation is especially appealing for modeling phenomena for which there is no simple analytical model, such as those governed by a succession of probability laws, or by specific detector or patient features. Simulations can also be extremely useful for determining the parameters of an analytical model that is then used to produce the SM. In that latter case, SM entries are not directly derived from simulations, but they are set via a model that is itself parameterized using simulations. To distinguish between these two approaches, we will call these latter SM as MC-driven-SM, while MC-SM will refer to matrices for which each entry is directly derived from simulations.

A SM can also be factorized into a product of independent submatrices, each describing an aspect of image formation (detector geometrical component, particle interactions within the object, particle interactions within the detector, positron range in PET, etc) [11]. This reduces the size of the matrices to be stored and allows for an independent computation of each contribution using the most appropriate model. In that approach, only one or some components can be calculated using MC simulations, while others can be accurately set analytically [12]. This decomposition of the SM will be called factorized SM in the following.

2.2. Effects modeled in simulation-based SM matrices

Simulations are used either to comprehensively model the detector response function (DRF), or to model the probability of particle interactions within the patient, or both.

In SPECT, Floyd et al initially modeled the scatter and attenuation occurring in the object or patient to derive an MC-SM (2D approach), already demonstrating the qualitative and quantitative gain brought by the method [5,6] in 2D. Then, the same group included the DRF in their model. In SPECT, the DRF depends on the distance between the source and the collimator. This distance dependence is due to solid angles defined by the collimator holes and results in a position dependent non-symmetry in the reconstructed image point response. Modeling this effect in the MC-SM led to an impressive improvement in the spatial resolution of the reconstructed images (FWHM of a line spread function reduced by a factor of ~2, [13]). This effect is now almost systematically compensated for in SPECT iterative reconstruction using an analytical (as opposed to a MC) model [14] but this early work demonstrated the importance of accounting for the DRF in SPECT reconstruction. The DRF consists of a geometric component, a septal penetration component and a collimator scatter component. The geometric response can be easily modeled analytically based on the detailed geometric specifications of the hole and septa of the collimator and is the dominating component for low energy radionuclides. The septal penetration and collimator scatter response are more difficult to model and should not be neglected for radionuclides emitting medium or high-energy photons, such as I-123, In-111, Ga-67 and I-131. Simulations are then extremely helpful to account for these components. They can be used to generate a table of DRF as a function of the distance between the source and the detector for a MC-driven-SM approach (eg, [15–18]) or to directly calculate the MC-SM entries (eg, [19]).

In PET, the very first attempts of MC-SM calculation included the detailed modeling of photon interactions within the detector (Compton scattering, detection efficiency, attenuation), the positron range and the non-collinearity of the annihilation photons, in 2D