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A dual modality ultrasound-gamma system: Monte Carlo simulations of the scintillation imager

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Abstract

The aim of the ECORAD collaboration is to develop a dual integrated compact and portable camera able to acquire ultrasound and scintigraphic images at the same time. In this work we present some simulated results of the scintigraphic part of the system. This camera consists of a rotating slant collimator with four segments connected to a planar LaBr3:Ce scintillator and to a PMT Hamamatsu Flat Panel H8500. Simulations are achieved by means of the GEANT4 program. The volumetric information is reconstructed from the planar images acquired at each position of the rotating collimator by means of a simple back-projection method. Results showed that the planar spatial resolution is better that the axial one. First preliminary results suggest that the detection limit of the camera is about 15:1 (in terms of Tumor/Background ratio) for a spherical tumor with 8 mm diameter located at 3 cm distance from the collimator. That confirms that with our approach it is feasible to developed a compact camera able to recover the 3D position of lesions located at small depths (up to some centimeters), without the need of rotating the camera around the body.

Keywords: Monte Carlo simulations; scintillating detectors; multi-modality systems; slant collimators.

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1. Introduction

Medical diagnoses commonly rely on the assessment of both the functional status and the anatomic condition of the patient. The aim of the ECORAD collaboration is to develop a dual integrated compact and portable camera able to acquire simultaneously ultrasound and scintigraphic images of the same portion of the body. Hence, a final volumetric image containing both functional and morphological information will be made available to the user. GEANT4 is a general purpose Monte Carlo program, well suited for transporting particles in various fields [1]. It also allows simulating the transport and boundary characteristics for optical photons generated by scintillating crystal. The entire system was simulated, starting from the transport of gamma photons within the phantom, and halting when the optical photons reach the photomultiplier tube. In recent years, we gained some experience in simulating experimental systems for medical imaging [2,3]. In this paper we used GEANT4 (version 4.9.0) to perform a preliminary evaluation of the scintigraphic part of the dual system that will be developed by the ECORAD collaboration.

2. Materials and methods

The simulated system is based on a four-segment slant-hole collimator. Slant collimators have the remarkable feature of being capable to provide a 3D image even with a stationary gamma camera. They have been recently used in gamma cameras for cardiac and breast imaging [4,5]. The simulated collimator has the following characteristics: 28 mm length, hexagonal holes with 1.8 mm diameter (the sensitivity is about 500 cpm/μCi), and slant angle of 30°. Each segment of the collimator is coupled to a compact gamma camera consisting on a multi anode PMT (Hamamatsu Flat Panel H8500), and a 50×50×4 mm^3 continuous $LaBr_3$:Ce crystal. The camera rotates around the vertical axis (*z*), thus providing several projections of the object. Actually, for each position of the camera, four projections are acquired, one for each segment of the camera. The Field of View (FOV) of the camera is determined by the volume intersected by all projections. The volume outside the camera is only partially recorded and details here will be imaged with less resolution and contrast.

We reconstructed the volumetric image by means of a simple back-projection method. The spatial resolutions in the plane parallel to the detector and in the axial direction perpendicular to that plane (*z* axis) are estimated by simulating point sources positioned at different locations. We calculated the spatial resolution as the Full Width at Half Maximum (FWHM) of the reconstructed Point Spread Function.

We have also simulated a phantom consisting of a cuboid made of soft-equivalent tissue with dimension 6×6×10 cm^3. In order to emulate a clinical examination, we simulated the number of photons emitted in a total imaging time of 10 minutes. We simulated 16 projections, corresponding to four different positions of the four-segment camera. Photons had energy equal to 140 keV, typical for the most common radioisotope used in SPECT applications (*i.e.* Tc^{99m}). The simulated background activity is fixed to 100 nCi/cc, and spherical tumors with a diameter of 8 mm and 10 mm were inserted within the phantom at a distance of 3 cm from the collimator. We simulated various Tumor/Background (T/B) ratios, ranging from 8:1 to 20:1. For each reconstruction of the phantom we calculated the Signal-to-Noise Ratio (SNR) on a Region of Interest (ROI) centered over the simulated tumor, in order to assess the quality of the reconstructed image.

3. Results

The reconstructed volumetric images have cubic voxels with a side of 1 mm. Figure 1 shows an example of the reconstruction of three point sources located on the central axis at different distances from the collimator. In this case, the three sources were simulated together, thus we can asses whether our system is able to discriminate sources positioned at different depths. The same picture illustrates both the reconstructed slices at the depth where the source is supposed to be located and a profile along the *z* axis. It is worth noting that the spatial resolution is better for sources close to the collimator, as expected. Further, we can state that the depth resolution of the system is accurate enough to distinguish point

sources spaced apart at a distance of 2 cm, at least for depths up to 5 cm from the camera.

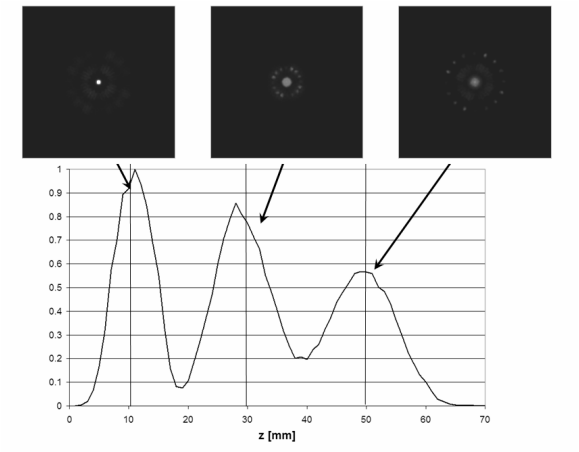

Figure 1: reconstruction of three point sources located on the z axis at a distance of 1 cm, 3 cm, and 5 cm from the collimator. Top: reconstructed slices at depth 1 cm, 3 cm, and 5 cm. Bottom: z profile of the reconstructed volume estimated on a central ROI.

In order to give a more complete assessment of the spatial resolution of the system across the entire FOV and also outside it, we simulated point sources located at different depths and positioned either along or outside the central axis at a certain distance from the central one. Figure 2 shows a plot of the FWHM of some of the simulated sources. The axial FWHM is worse (about double) than the planar one. Moreover, the spatial resolution clearly gets worse for points outside the FOV.

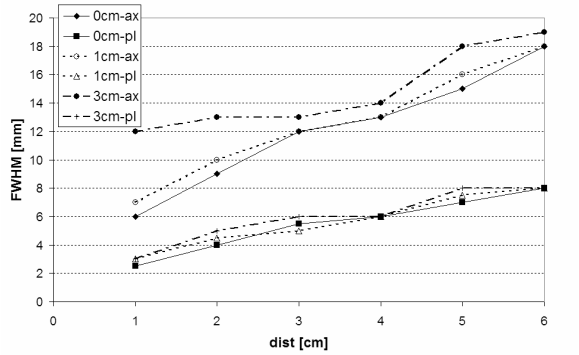

Figure 2: axial and planar spatial resolutions (FWHM) of the reconstructed point sources, as a function of the distance from the collimator. Sources are located on the central axis (0 cm), and on two lateral positions with a distance of 1 cm and 3 cm from the central axis.

Table 1 shows some SNR values calculated for the 8 mm and 10 mm tumors with different radioactivity concentrations located at 3 cm from the collimator. By assuming the standard detection limit of SNR equal to 5, we can note that the 8 mm tumor is visible for T/B ratios greater than 15:1, whereas the 10 mm tumor is perceptible also for smaller T/B values.

Table 1: SNR values for 8 mm and 10 mm tumors located at 3 cm from the collimator for different T/B values.

T/B	**8:1**	**10:1**	**15:1**	**20:1**
8 mm	3.0	3.2	5.4	6.7
10 mm	3.7	5.5	8.6	9.2

4. Conclusions

First preliminary results suggest that for a 10 minutes examination the visibility limit of the proposed camera is about 15:1 for the 8 mm diameter spherical tumor and 10:1 for the 10 mm one, both located at 3 cm from collimator. Thanks to the slant collimators, we thus demonstrate the feasibility of detecting small tumors with a camera able to trace the depth of a lesion, without the need of rotating it around the body. We believe that this remarkable feature represents an important advance for the development of compact and portable devices dedicated to the imaging of lesions located at small depths (up to a few centimeters).

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