VERSATILE COMPTON CAMERA FOR HIGH-ENERGY GAMMA RAYS: MONTE CARLO COMPARISON WITH ANGER CAMERA FOR MEDICAL IMAGING*

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Single Photon Emission Computed Tomography (SPECT) is at present one of the major techniques for non-invasive diagnostics in nuclear medicine. Almost the whole clinical routine is based on collimated cameras, originally proposed by Anger. The presence of mechanical collimation limits detection efficiency and energy acceptance. The application of Compton cameras for SPECT could allow to overcome these limitations. In this study, we propose to compare our Compton camera prototype to a commercial Anger device, the GE Healthcare Infinia system, through Monte Carlo simulations (GATE v7.1 and Geant4 v9.6 respectively). Given the possible introduction of new radioemitters at higher energies allowed by the Compton camera detection principle, the detectors are exposed to point-like sources at increasing gamma energies. The detector performances are studied in terms of radial event distribution, detection efficiency and final image, obtained by gamma transmission analysis for the Anger system, and with an iterative LM-MLEM algorithm for the Compton reconstruction. Preliminary results show for the Compton camera a detection efficiency increased of a factor greater than an order of magnitude, associated with an enhanced spatial resolution for high energies. We then discuss the proven advantages of the Compton camera application with particular focus on dose delivered to the patient, examination time and spatial uncertainties.

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1. Introduction

In most of the SPECT clinical cases, a radiotracer is injected in the patient and the emitted γ-rays are collected by Anger-based detectors [1], composed by a mechanical collimator coupled to a scintillator with position-sensitive readout photomultipliers.

The presence of physical collimation system leads to a forced trade-off between sensitivity and spatial resolution, and limits the exploitable gamma energy range. The use of radioemitters at high energy (> 500 keV) has already been proposed for SPECT applications in previous studies [2], and several radioisotopes are already available. The standard SPECT camera limitations can be overcome by the introduction of an "electronic collimation", where the emitted photons are tracked and the emission point is reconstructed via Compton scattering. A Compton camera is generally composed of two detector stages: a scatterer and an absorber. The position and energy information given by the two detectors components allow one to limit via analytic or iterative algorithms the emission point on the surface of a cone thanks to the Compton scattering relation. The radiotracer distribution with a single compact detector is then reconstructed through the intersection of several cone surfaces.

In the present work, we investigate the possible application of the Compton camera prototype under development by the CLaRyS Collaboration between five laboratories in France [3] in SPECT. Moreover, we compare our detector to the Infinia Anger camera delivered by GE Healthcare [4]. The detector performances are compared in terms of efficiency and spatial response with the exposure to mono-energetic point-like radioactive sources at different energies, ranging from 245 keV to 2.614 MeV.

In a previous article [5], we have performed a detailed comparison between the Compton and Anger detection devices in terms of efficiency and spatial resolution. However, we want to stress in the present one the choice of the detection material, and the subsequent influence of the Doppler broadening, that were not described yet.

2. Methods

2.1. Detector settings

The two cameras are simulated with GATE v7.1 (Anger) and Geant4 v9.6 (Compton).

The Infinia camera is modeled according to the provider specifications [4] and it is equipped with a High Energy General Purpose (HEGP) collimator.

The Compton camera reproduces the design presented in [3]. The absorber total transverse size is adapted to be as close as possible to the Anger

camera, according to the single block size, with a matrix of 8×6 BGO blocks. In addition to this configuration, a more compact one has been tested with an absorber of reduced size, with a matrix of 3×3 blocks (10.5×10.5 cm^2).

Concerning the scatterer part, the position of each interaction is set to the center of the strip where it is recorded in each detection plane (the strip pitch is 1.4 mm), and at the center of the involved detector slab in the perpedicular direction. The time resolution has been set to 20.0 ns FWHM, while the energy resolution is set to $\sigma_E = 2$ keV.

The energy and time resolution of the BGO blocks are set to 21% FWHM and 3.0 ns FWHM respectively. Each block surface is streaked with a 8×8 pixel matrix, 4.4 mm side, not reproduced in the simulation code. Each interaction is then spatially assigned to the center of the pixel where it is localized at the analysis stage, while the interaction depth is set to the center of the involved block.

2.2. Data acquisition and analysis

The two detectors are exposed to 13 mono-energetic sources, at 10 cm distance from the collimator entrance and from the first silicon plane, at increasing energies from actual sources already used in our previous simulation work [5].

A source activity of 200 MBq (consistent with actual clinical practice) has been selected for the comparison study [5].

The Compton camera events composed by one interaction in a single scatterer plane and one or several interactions in a single absorber block are selected and then reconstructed with a list-mode MLEM iterative algorithm [6] (see figure 1).
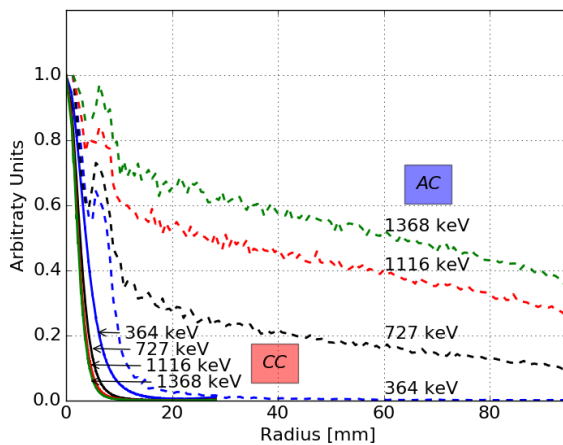

Fig. 1. (Color online) Overlap of the normalized radial distributions for four selected source energies for Compton (CC — solid lines) and Anger (AC — dashed lines) camera.

The Compton camera design has been checked for what concerns the main instrumental development challenge: the scatterer energy resolution. The impact of Doppler broadening must be minimized: it increases with the detector material Z, so that silicon (lowest Z available detecting material) should be the best choice for this section. The cadmium telluride (CdTe) has been tested as an alternative material. For both materials, the Doppler broadening effect has been disabled at the simulation stage to compare the results and estimate the spatial degradation.

A simple analysis method based on linear fit on the raw event radial distribution has been applied to the Anger camera data for background rejection. The events are then selected by subtracting the fit result to the raw radial distribution.

The comparison between the two detectors is based on three customized figures of merit relying on spatial performance, detection efficiency and event selection:

— the spatial resolution: defined as the standard deviation of the radial distribution of the selected events after background rejection (Anger) or after MLEM reconstruction (Compton);

— the detection efficiency: defined as the ratio between the selected (Anger) or reconstructed (Compton) events and the total number of simulated primaries;

— the signal-to-noise ratio: defined as the ratio between the selected (Anger) or reconstructed (Compton) events and the total detected events (Anger) or detected coincidences (Compton).

The three described parameters are studied as a function of the gamma energy leading to a direct comparison of the two detectors performances, including the two Compton absorber configurations already explained in Section 2.1.

3. Results and conclusions

Figure 2 shows the standard deviation of the radial distribution of reconstructed events in the Compton camera for the two studied scatterer materials, silicon and CdTe. The results confirm the expectations: the camera with the silicon scatterer outperforms the one with CdTe scatterer by more than 1.8 mm in the whole studied energy range. For the silicon scatterer, the Doppler blurring degrades the spatial resolution by more than 3 mm at low energy, while the contribution is reduced at higher energy. For what concerns the CdTe scatterer, the higher Z leads to an increased Doppler degradation by more than 5 mm at low energy, as before reduced at higher energy. Moreover, these results allow one to verify that the Doppler broadening is the main source of spatial response degradation.
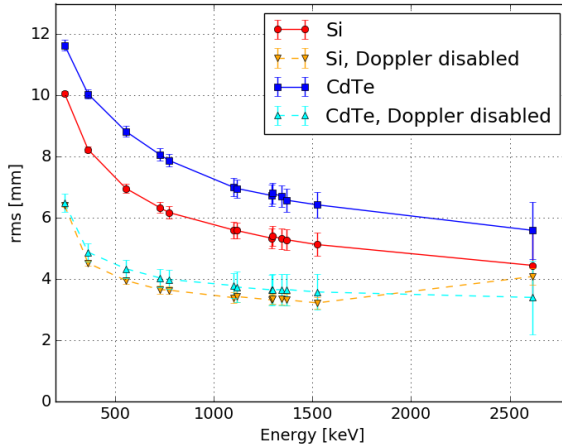
Fig. 2. Standard deviation of the radial event distribution as a function of the source energy for two scatterer materials, Si and CdTe, with and without Doppler broadening effect. Source activity = 200 MBq, scatterer detector σ_E = 2 keV.

Figures 3 and 4 present the results of the Compton camera comparison to the commercial Anger camera Infinia by General Electrics. The results presented in [5] are here reported for comparison to the ones obtained with Compton camera with reduced absorber, 10.5 cm side. At the expense of a reduction in the detection efficiency with respect to the wider Compton camera prototype (factor > 2 at low energy, reduced at higher energies), an improvement in the spatial resolution is detected, as well as an enhanced signal-over-background ratio. This is due to the event selection already performed at the detector level: a small absorber area determines a selection of Compton events at small angles, resulting in a better spatial resolution. Moreover, the amount of random coincidences is reduced, so that the signal-over-background ratio is improved. The efficiency gain with respect to the Anger camera is of a factor close to 10 for the whole energy range (notice the two different scales in figure 1), while the spatial resolution is advantageous for energies above approximately 400 keV.

These results illustrate the potential of the Compton camera for the application in nuclear medicine examination, opening new possibilities for the clinical implementation. The enhanced detection efficiency in parallel with comparable spatial performances paves the way to the extensive usage of less active sources, or alternatively allows a substantial reduction of examination time: as a result, the dose delivered to the patient would be reduced. In addition, the possible introduction of sources with higher primary emission energy (without a reduction in detection efficiency which is forced for the Anger systems) will reduce the effect of photon attenuation in the patient
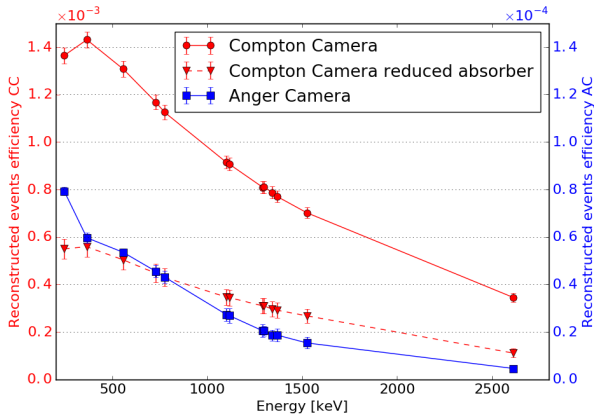
Fig. 3. Detection efficiency as a function of the source energy. Source activity = 200 MBq, Compton camera Silicon detector $\sigma_E = 2$ keV. Left scale for Compton camera, right scale for Anger camera.
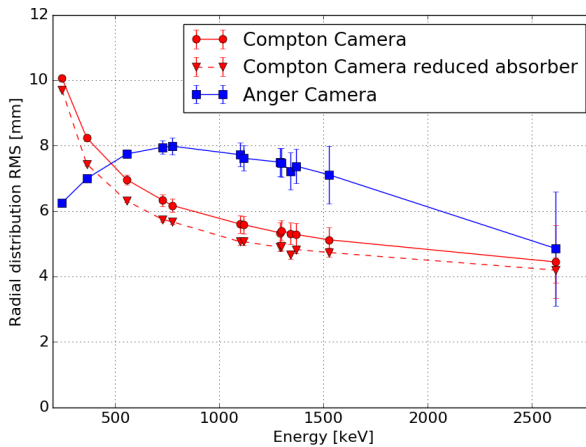

Fig. 4. Standard deviation of the radial event distributions as a function of the source energy. Source activity = 200 MBq, Compton camera Silicon detector $\sigma_E = 2$ keV.

(not studied in this simulation work), improving by definition the spatial information and further reducing the effective dose delivered to the patient. Finally, the already underlined gain in detection efficiency and the enhanced spatial resolution at high energy, in parallel to the wider gamma energy acceptance, make the Compton camera suitable for targeted radionuclide therapy.

REFERENCES

[1] H.O. Anger, *Rev. Sci. Instrum.* **29**, 27 (1958).

[2] T.C. Nurdan, K. Nurdan, A.B. Brill, A.H. Walenta, *JINST* **10**, C07018 (2015).

[3] J. Krimmer *et al.*, *Nucl. Instrum. Methods Phys. Res. A* **787**, 98 (2015).

[4] GE Healthcare, Infinia, release 2.5, 2006.

[5] M. Fontana *et al.*, Compton Camera Study for Nuclear Medicine and Benchmark with Anger System, submitted to *Phys. Med. Biol.*

[6] V. Maxim *et al.*, *Phys. Med. Biol.* **61**, 243 (2016).