Evaluation of Compton gamma camera prototype based on pixelated CdTe detectors

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ABSTRACT: A proposed Compton camera prototype based on pixelated CdTe is simulated and evaluated in order to establish its feasibility and expected performance in real laboratory tests. The system is based on module units containing a $2 \times 4$ array of square CdTe detectors of $10 \times 10 \text{mm}^2$ area and 2 mm thickness. The detectors are pixelated and stacked forming a 3D detector with voxel sizes of $2 \times 1 \times 2 \text{mm}^3$. The camera performance is simulated with Geant4-based Architecture for Medicine-Oriented Simulations (GAMOS) and the Origin Ensemble (OE) algorithm is used for the image reconstruction. The simulation shows that the camera can operate with up to $10^4 \text{ Bq}$ source activities with equal efficiency and is completely saturated at $10^9 \text{ Bq}$. The efficiency of the system is evaluated using a simulated $^{18}\text{F}$ point source phantom in the center of the Field-of-View (FOV) achieving an intrinsic efficiency of 0.4 counts per second per kilobecquerel. The spatial resolution measured from the point spread function (PSF) shows a FWHM of 1.5 mm along the direction perpendicular to the scatterer, making it possible to distinguish two points at 3 mm separation with a peak-to-valley ratio of 8.

KEYWORDS: Compton imaging; Detector modelling and simulations I (interaction of radiation with matter, interaction of photons with matter, interaction of hadrons with matter, etc); Solid state detectors

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1 Introduction

Single Photon Emission Tomography (SPECT) is a widely used diagnostic tool in nuclear medicine due to its low cost compared to other techniques, the ability to use single photon emitter sources which broadens the range of radio-tracers available, and the planar design which does not impose limits on the patient size. SPECT uses a mechanical collimator in combination with a position sensitive detector and has intrinsic drawbacks coming from the mechanical collimation. Only a small fraction of the gammas can pass through the collimator, resulting in an inherently low efficiency and inverse relation between efficiency and spatial resolution. Moreover, a thicker collimator must be used for high energies due to the increasing penetrating power of the gamma.

Compton camera has been proposed [1, 2] to overcome these limitations and the evaluation of several prototypes has been reported [3–7]. A Compton camera consists of two detectors working in coincidence. In a coincident event the gamma photon undergoes a Compton interaction in the first detector (the scatterer) and is then completely absorbed in the second detector (the absorber). If we assume that the electron is at rest when the Compton interaction takes place, we can relate the Compton angle \( \theta \) with the deposited energy in the scatterer \( E_{\text{scatt}} \) with the following equation:

\[
\cos \theta = 1 - m_e c^2 \left[ \frac{1}{E_\gamma - E_{\text{scatt}}} - \frac{1}{E_\gamma} \right],
\]

(1.1)

where \( m_e c^2 \) is the electron rest energy and \( E_\gamma \) is the gamma energy before the scattering. With the Compton angle and the hit positions in both the scatterer and the absorber, one can reconstruct a cone on which the emission point must lie. Because the electron in the scatter interaction has momentum, equation (1.1) is not completely valid. This introduces an error in the computation of the angle which is called the Doppler broadening effect [8].

Compared with traditional SPECT systems the Compton camera yields higher efficiency as every gamma reaching the scatterer has a good probability to undergo a Compton scattering interaction. Given a fixed configuration, it works in a wider range of energies as the change in energy
will only affect the interaction probability in both detectors, the scatterer and the absorber. Compton cameras can work with radio-tracers of different energies simultaneously [4]. As the Compton camera works with cones rather than projections it has the potential to produce three dimensional images without the need of moving the device.

Figure 1. Schematic of the module unit. The scatterer is made with one single module while the absorber is the result of the stacking of 10 modules. The voxel density is 182 voxels/cm$^3$.

The limitations in the image spatial resolution in Compton cameras come from the errors in the cone reconstruction. The accuracy of the reconstruction of the cone depends on the camera’s spatial and energy resolution in both interactions, and the Doppler broadening effect. In the proposed prototype the spatial resolution is limited mainly by the voxel size. The detector energy resolution and the Doppler broadening effect affects directly the computation of the Compton angle through equation (1.1).

In the past we have modeled and evaluated a Compton camera based on pixelated Si/CdTe detectors [9] making use of the technology that we are developing for PET [10] and PEM [11]. In this new study we want to evaluate a prototype Compton camera based on pixelated CdTe detectors. The simulations studies presented serve as proof of concept to assess the viability of constructing such prototype.

## 2 Specifications and simulation setup

### Compton camera specifications

The scatterer and absorber of the Compton camera prototype are built up from module detector units. An individual module contains an array of $2 \times 4$ CdTe detectors. The dimension of the CdTe detectors is $10 \times 10 \times 2$ mm$^3$. The CdTe detectors are sitting on a 300 µm SiO$_2$ glass substrate placed upon a 100 µm flex-ridge kapton layer. The CdTe detectors are pixelated with a $2 \times 1 \times 2$ mm$^3$ voxel size (see table 1).

The absorber is made by stacking 10 modules. In this way, we create a 4000 channels 3D voxel array with a density of 182 voxels/cm$^3$. The absorber’s dimensions are $2 \times 2 \times 4$ cm$^3$. The scatterer consists of just one module unit. The scatterer to absorber distance is set to 5 cm. The absorber is vertically stacked in such a way that the detector can easily be made bigger by simply adding more modules (see figure 1).

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**Table 1. Camera Prototype Parameters.**

| Parameter                  | Value                        |
|----------------------------|------------------------------|
| Detector Energy Resolution | 1.57% FWHM at 511 keV        |
| Measuring Time             | 500 ns                       |
| Dead Time                  | 51.2 µs                      |
| Time Resolution            | 100 ns                       |
| Trigger Threshold          | 20 keV                       |
| System Area                | 20 mm $\times$ 40 mm         |
| Scatterer Thickness        | 2 mm                         |
| Absorber Thickness         | 20 mm                        |
| Scatterer to Absorber Distance | 50 mm                     |
The FWHM energy resolution for 2 mm thick CdTe at 511 keV incident gamma energy is 1.57% [12] at room temperature.

The 400 independent channels per module are read out by four VATAGP7 ASICs [13]. VATAGP7 is a 128-channel charge-efficiency amplifier and self-triggered.

The absorber detector, with 4000 channels, and the scatterer detector, with 400 channels, are treated as independent detectors with the data collected in list mode and processed offline.

**Data analysis and image reconstruction.** The full camera is simulated with Geant4-based Architecture for Medicine-Oriented Simulations (GAMOS) [14]. GAMOS allows us the simulation of the full detector including passive material and all the processes including Doppler broadening effect.

![Figure 2](image.png)

*Figure 2.* The camera is rotated creating a cylindrical FOV of 25 mm radius and 40 millimeters length.

The data obtained from the GAMOS simulation is processed in order to replicate the acquisition process which takes place in the real detector. The acquisition process begins with the activation of a voxel which occurs when the deposited energy exceeds the 20 keV trigger threshold. The trigger defines the timestamp of the hit after which 500 µs measuring time and 25 µs dead time is applied. All energies deposited within the same voxel and within the measuring time contribute to the total energy of a single hit. Energies deposited within the dead time in any of the voxels corresponding to the chip in which the trigger took place are lost.

After the data acquisition, the coincidence selection algorithm processes the data to group consecutive hits inside a 100 ns coincidence time window. Events with a single hit in the scatterer and a single hit in the absorber are selected and from these only the ones in which the combined energy of the hits is 511 keV with a ±1.57% tolerance are taken as coincidence events.

The image reconstruction is performed using the OE algorithm [15]. With OE an estimation of the origins of the events based on its density in the FOV is obtained. Initially a random position for each event is selected on the corresponding reconstructed cone. Then, in each OE iteration a new random position on the cone is calculated for every single event and is accepted or rejected with a probability that depends on the event densities in the voxels corresponding to the old and new position.

The output of the algorithm is a 3D map of the reconstructed activity of the source. Due to the stochastic nature of OE several independent trials must be performed and averaged. Each trial consists of one single run of the algorithm but with a different random seed. In all the images presented 50 trials and a minimum of 1000 iterations per trial are performed. The performance of OE on PET, PEM, and Compton camera devices has been evaluated [16], showing its feasibility for devices with a large number and high density of channels.

The distance of the FOV center to the scatterer edge is 25 mm in all the simulations. The distance is set to be comparable with the shorter length of the FOV(20 mm) thus the angle covered
by the detector is small. The cone density on the solid angle covered by the detector is high enough to create artifacts in the reconstructed image given that OE uses this density to accept or reject new locations for the event position in each iteration. To increase the covered angle we rotate the detector at constant time intervals covering the 360 degree circumference in twenty steps of 18 degrees each (see figure 2). The 50 mm aperture of the cylindrical FOV is enough for small animal test applications.

3 Results

3.1 System efficiency

In order to measure the source activity range in which the system can operate and evaluate the efficiency of the camera, different simulations have been performed. In all the simulations a point-like source phantom based on NEMA NU-4 2008 [17] is used. The phantom consists of a cube of 10 mm side of acrylic material with a 0.1 mm radius sphere in its center. The sphere is filled with active water containing $^{18}$F isotope.

In figure 3 we compute the absolute efficiency of the Compton camera (i.e., the number of coincidence events divided by the number of emitted gammas in all the solid angle) in counts per second per kBq for increasing activities. The events are classified as “true”, “random” (i.e., coming from two unrelated gammas), “back-scattering” (i.e., gammas that make a high energetic Compton interaction in the absorber and go to the scatterer where it is absorbed) or “other”. The main contribution to the “other” category are events with more than one hit in scatterer or absorber but in which the extra hits are below trigger threshold. The secondary contribution comes from photoelectric interactions in the scatterer followed by the emission of a bremsstrahlung photon, with an energy above the trigger threshold, which undergo a photoelectric interaction in the absorber detector. One can see that the count rate begins to drop at $10^5$ Bq and the camera is completely saturated at $10^9$ Bq. The absolute efficiency is 0.03 cps/kBq. The fraction of true events over the total number of coincidence events is 81% in the entire non-saturated range. For all the following simulations the activity is set at $10^4$ Bq where the saturation effects are negligible.

To compare the obtained efficiency with commercially available SPECT systems, one needs to compute the intrinsic efficiency (i.e., the number of coincidence events divided by the gammas that reach the detector). For a point-like source in the center of the FOV, the absolute efficiency
$\epsilon_{\text{abs}}$ and the intrinsic efficiency $\epsilon_{\text{int}}$ are related by the following formula:

$$
\epsilon_{\text{int}} = \epsilon_{\text{abs}} \cdot \left(\frac{4\pi}{\Omega}\right),
$$

where $\Omega$ is the solid angle covered by the detector. The obtained intrinsic efficiency of the Compton camera is 0.4 cps/kBq which is higher than any commercially available scintillator based SPECT system using a ultra-high energy collimator.

Approximately 10% of the total number of events are due to “back-scattering”. These events contribute to the noise as they will yield a wrong Compton angle given that the real Compton interaction happens in the absorber. Due to the small size of the detectors compared to the scatterer to absorber distance the “back-scattering” events are distributed in a well defined peak on the energy spectrum. The simplest way to eliminate the back-scattered events is to fit the “back-scattering” peak and cut all the events in a fixed range around the peak mean, as shown in figure 3.

As has been shown [18], silicon is a superior choice as scatterer detector material for its reduced Doppler broadening and its bigger Compton cross section for low energy gammas, as emitted from $^{99m}$Tc. However, for 511 keV gammas a thicker Si scatterer is needed. In figure 4 the absolute efficiency of the Compton camera is computed when CdTe is substituted with Si in the scatterer detector. At least four module units containing 2 mm thick Si detectors must be stacked above each other to match the efficiency of a single layer of CdTe for 511 keV gammas.

### 3.2 Image reconstruction

**Point-like source.** In order to characterize the image spatial resolution we can expect from the prototype, an image of the point-like source phantom in the center of FOV is reconstructed from 10k coincidence events.

Here and in the next results, the image FOV is $10 \times 10 \times 10$ mm$^3$ segmented in cubic bins with sides of 0.5 mm. By rotating the Compton camera around the z-axis, the FWHM for the y and x directions will average out in between the FWHM for the radial and the axial directions. The FWHM in the z-direction will always correspond to the axial direction with respect to the Compton camera planes. As the FWHM for the radial direction (perpendicular to the Compton camera planes) is worse than in the axial direction, the FWHM for y and x is always worse than the FWHM for z. The width of the Point Spread Function (PSF) is computed through the FWHM of the line profiles over the...
radial and axial directions. We obtain a FWHM of 2.2 mm along radial direction and 1.5 mm along axial direction (see figure 5).

Comparable spatial resolution is obtained with Compton camera prototypes using Double-sided Silicon Strip Detectors (DSSD) as scatterer [19].

**Two point-like sources.** To test the resolving power of the prototype we simulated two point-like sources with 3 mm separation. In the simulation we use a modified version of the phantom used for the point-like source. It consists of a rectangular acrylic block with $10 \times 10 \text{mm}^2$ base area and 13 mm height. Two 0.1 mm spheres are placed at 1.5 mm from the center of the acrylic block along the longest direction. The phantom is placed with respect to the camera in such a way that the sources are aligned with the axial direction of the cylindrical FOV.

The image is reconstructed using 20k coincidence events. The two point-like sources can be clearly distinguished with a peak-to-valley ratio of 8 as seen in figure 6. The PSF for each of the points remains equal at 1.5 mm FWHM.

**Point-like sources with different activities.** The ability to accurately distinguish different activities is important to differentiate the signal from background activity and to distinguish different tissues in real medical applications. In order to quantify the accuracy of our Compton camera when reconstructing different activities, we used a phantom with the same dimensions of the one used with two point-like sources, but now it contains three point-like sources with 5 mm separation and activity ratios of 1:2:3 times 100 kBq. In figure 7 the image reconstructed from 30k coincidence events is shown.

To measure activity ratios from the image we fit each one of the three peaks and integrate all the events in a region of five sigma around the mean value. Normalizing the result to the lowest peak we obtain a ratio between the activities of 1:2.4:3.2. Hence the activity ratio is measured with accuracy better than 80%.

**Point-like source on cube vertices.** To show the full 3D imaging capabilities of the Compton camera we simulate an array of point-like sources arranged in the vertices of a cube. The phantom used in the simulation is again a modified version of the one used for the point-like source. It consists of a cube with sides of 15 mm length made of acrylic material which contains eight point-like sources of radio-tracers placed at the vertices of another cube with sides of 10 mm, see figure 8.
The image reconstructed using 80k coincidence events of a three dimensional array of eight points in the vertices of a cube is shown in figure 8. Two slices of the fully reconstructed 3D image with different plane orientation are shown. The FWHM is consistent with the PSF analysis and the points are clearly distinguishable and with compatible intensities.

4 Conclusions

A simulation study has been presented for Compton gamma camera, under construction, that has pixelated \textit{CdTe} detectors, stacked above each other, to form a true 3D detector.

The simulation results, including all contributions to the detection resolution and the Doppler broadening effect, are presented here. Using the OE imaging reconstruction algorithm one can obtain 3D images with an excellent spatial resolution of 2 mm FWHM, from phantoms with multiple sources with varying activities. The simulation results are promising and encouraging to set the case to construct high density stacked pixelated CdTe detectors. The presented prototype serves as a proof of concept to construct a detector with similar design but with a larger area, e.g 30 cm $\times$ 30 cm, that matches the common needs of the nuclear medical imaging practice.

An implementation of a more sophisticated image reconstruction algorithm (e.g. LM-OSEM) for more complex image phantoms, is presented in [20].

Acknowledgments

This work has been supported by FP7-ERC-Advanced Grant #250207.

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