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Application of silicon strip detectors to small-animal imaging

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Abstract

We have investigated the design of a high-resolution nuclear medicine system for in vivo small-animal imaging. The system utilizes a multiple-pinhole collimator in conjunction with silicon double-sided strip detectors in a synthetic collimator configuration. It is designed for detection of the low-energy photons (27–35 keV) from the decay of ¹²⁵I or ¹²³I. We discuss elements of the system and present simulations conducted thus far.

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1. Introduction

There has been rapid growth recently in demand for in vivo small-animal imaging capabilities. Large numbers of animal models for human diseases have been developed, and techniques for probing gene expression are quickly evolving. Moreover, in vivo imaging has many scientific, economic, and ethical advantages over other well-established biological imaging techniques, such as autoradiography. While nuclear medicine techniques can be applied to small animals, standard clinical systems generally are incapable of meeting the resolution and sensitivity demands encountered in small-animal imaging [1]. The system that we propose is a compact device well suited to small-animal imaging demands and one that offers the potential for unprecedented resolution.

2. Silicon double-sided strip detectors

The development of silicon double-sided strips detectors (DSSDs) as radiation detectors has been driven by their use in high-energy physics, where they have found widespread use as vertex detectors in collider experiments. The readout electrodes (strips) on the front and back of DSSDs are orthogonal to one another, making it possible to determine the (x, y) coordinates of a photon interaction from the strips on each side that had a signal induced. Several companies manufacture custom-designed silicon DSSDs with considerable flexibility in detector area (up to $\approx 6.4 \times 6.4$ cm² using a 4 in. wafer), strip pitch (from 20 μ m to the full detector size), and thickness (300 μ m to 2 mm). The large number of readout channels incurred in using DSSDs necessitates the use of application-specific integrated circuits (ASICs) for the front-end electronics.

The application of silicon strip detectors to biomedical imaging has primarily been focused on

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the field of digital radiography [2,3]. In nuclear medicine, silicon strip detectors have been incorporated into a Compton-scatter imaging device [4]. The main perceived limitation of silicon detectors for nuclear medicine imaging applications is that the low Z -value means that they offer modest detection efficiency relative to other semiconductors (Ge, CdZnTe) or scintillators (NaI), particularly for the gamma-ray energies of typical interest in nuclear medicine (i.e., 140 keV for ^{99m}Tc). However, at lower energies silicon has reasonable detection efficiency (32% at 27 keV in 1-mm-thick detector). The overall efficiency can be further enhanced by stacking detectors (an additional 1 mm detector increases the total efficiency to 53% at 27 keV). This relative balance in count rate for stacked detectors actually is a significant advantage of using silicon detectors in a synthetic collimator configuration (see Section 4).

3. Radioiodination

Two radioisotopes with low-energy emissions have been identified as well matched for imaging with our silicon-based system. The first, ^{125}I , is used widely in molecular biology research because of the ease with which a variety of biological molecules can be iodinated and because its half-life (≈ 60 days) is convenient for radio-labeling work. The decay of ^{125}I is via electron capture with the emission of a 35 keV gamma-ray. Several K X-rays (27–32 keV) from the ^{125}Te daughter can accompany the decay, resulting in an average of 1.5 photons emerging for each ^{125}I decay. Another iodine isotope, ^{123}I , is already utilized in nuclear medicine, owing to the suitability of its half-life (13.2 h) for human use, and its 159 keV gamma-ray, which is well-matched to many clinical gamma cameras. Like ^{125}I , the decay of ^{123}I also results in the emission of K X-rays in the 27–32 keV energy range, with the total X-ray emission probability essentially equal to that of the 159 keV emission.

4. The synthetic collimator

Pinhole collimation for SPECT imaging of small animals has been utilized widely due to the ability

to get high-resolution images by magnifying the small-field-of-view object onto a large area detector (often a clinical gamma-camera) [5]. A substantial increase in sensitivity can be attained by using a multiple-pinhole collimator; however, multiple-pinhole collimators introduce the problem of multiplexing. Multiplexing occurs when projections through different pinholes overlap on the detector. A photon hitting a given detector pixel will have reduced information as it could have come through any of two or more pinholes.

Members of our group have shown that the problem of multiplexing can be overcome with a system we refer to as the synthetic collimator [6]. The idea is to collect multiple-pinhole projection data using a number of pinhole-detector distances, as shown in Fig. 1. Since the amount of overlap varies as a function of this distance, the data contain information on how the projections are multiplexed and on how to remove the multiplexing effects. Ideally, low-multiplexed data from small pinhole-detector distances and high-resolution data from larger pinhole-detector distances would be collected. The synthetic-collimator concept is to then use an algorithm such as maximum likelihood-expectation maximization (ML-EM) to combine these in a manner that leads to a nonmultiplexed, high-resolution reconstruction. Computer simulations have demonstrated potential advantages of synthetic collimation over parallel-hole collimation for 2D planar imaging and have shown further advantages exist when 3D rather than 2D images are reconstructed [7]. The

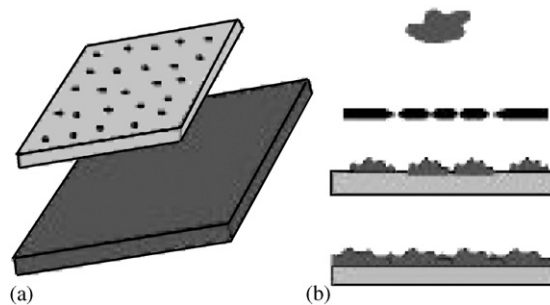


Fig. 1. (a) A three-dimensional view of the synthetic collimator, and (b) a schematic view of images collected at two different pinhole-detector distances, showing the tradeoff between magnification and multiplexing.

reconstruction of 3D images is possible even when data from only one collimator–object angle are collected.

One way to obtain synthetic-collimator data is to mechanically vary the distance between the multiple-pinhole collimator and the detector. Another possibility is to stack several detectors behind a single collimator. Stacking silicon detectors yields higher overall sensitivity, reducing the time needed for imaging.

5. Simulations

A number of simulations have been performed to study various aspects of the proposed imaging system. One important factor in dealing with low-energy photons is that, even with excellent detector energy resolution, scatter rejection is essentially impossible. A 27 keV photon loses a maximum of 2.6 keV in a Compton scatter. It is essential, therefore, to include the effects of scatter in studying the properties of any low-energy imaging system. Besides the complications of in-body scattering, because we envision using multiple layers of silicon in the imaging system, we also must worry about photons that Compton scatter in one layer of the system and then deposit their full energy in a different detector layer. As a first check of the magnitude of events of this type, we simulated a beam of photons entering a three-

layered detector system at normal incidence using the EGSnrc Monte Carlo code [8]. Each detector was taken to be 1 mm thick, and the distribution of photon energies was according to the decay scheme of ^{125}I . For a trigger threshold of 20 keV, the total detection efficiency was found to be 64%, with less than 3% of the detected photons undergoing a Compton scattering in one layer before depositing greater than 20 keV in another.

A preliminary study of the imaging capabilities of a single detector with a multiple-pinhole collimator has also been carried out. Two-millimeter slices from the digital “mouse” phantom used for this experiment are shown in Fig. 2. The objects in the phantom were a sphere of 2.5 mm radius containing a 1.5 mm hollow center (simulating a tumor with necrotic center) and a bright ellipsoid of length 6 mm and diameter 4 mm. The mouse body was modeled as a 12.5-mm-radius cylinder of water with a background specific activity of 5% relative to the objects. In order to simulate out-of-field activity, the cylinder was extended beyond the field of view shown in Fig. 2, with out-of-field length equal to the in-field length and radiotracer concentration equal to half the specific activity of the in-field portion. Four hundred million 27 keV photons were generated using the EGSnrc Monte Carlo code. This code not only accounted for absorption of photons in water, but also yielded an ensemble that included photons that had undergone scatterings (both

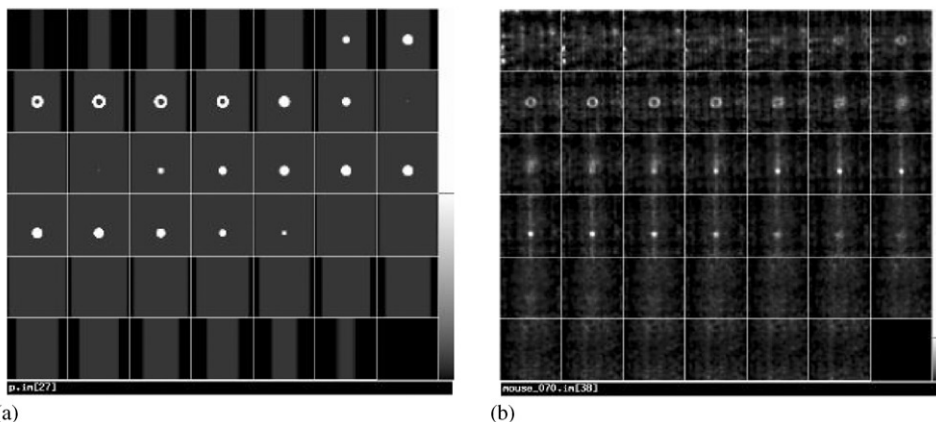


Fig. 2. (a) The “mouse” phantom used for the imaging simulation, shown in 2 mm slices. (b) Transverse image slices reconstructed from the simulated projection data.

Compton and Rayleigh) in the water. The photons that emerged were raytraced to the collimator plane, and those that passed through a pinhole were tracked to a pixel of the detector. The number of initial photons was determined by computational considerations, rather than expected activity and imaging times, and corresponded to roughly 30 s imaging time with 1 mCi of activity and a detector efficiency of 33%.

The photons were focused onto the detector by a 25-pin-hole collimator with the 1.0-mm-diameter pinholes randomly jittered about a regular 5×5 grid. The jittering was performed in an attempt to eliminate possible artifacts from redundant sampling. The collimator was placed 10 mm below and parallel to the upper left phantom slice in Fig. 2, and the collimator–detector distance was 22 mm. The detector had 320×320 0.2 mm pixels for a total surface area of $64 \text{ mm} \times 64 \text{ mm}$. This limited-angle tomography configuration subtended an angular range of $\sim 70^\circ$. Detector effects such as depth of interaction were not simulated.

The projection image contained approximately 700,000 total counts, indicating a collimator sensitivity (excluding detector efficiency) of 1.7×10^{-3} for our phantom. Reconstruction was performed using the ML-EM algorithm with no filtering or acceleration. The 2D slices of the reconstructed image (70 ML-EM iterations) are given in Fig. 2. Clearly seen is the “tumor” as well as the “necrotic center”. Less well defined is the ellipse, which lies farther from the detector.

6. Future work

More extensive simulations are planned to probe the capabilities of the proposed imaging system and address issues important to the design. The first step will be to combine simulation of photon interactions in the silicon detectors with the simulation of radiation transmission from the phantom through the collimator. Of particular interest is the effect of depth of interaction in the silicon detectors on the imaging resolution. We also plan to obtain an “off-the-shelf” detector system, consisting of a silicon DSSD and readout

electronics, for use in a prototype, single-layer imaging system to demonstrate the feasibility of the basic scheme. The synthetic collimator aspect will be tested by mechanically varying the collimator–detector distance.

7. Conclusions

We have developed a plan for a small-animal imager that incorporates silicon strip detectors in a synthetic collimator configuration. Silicon DSSDs offer position resolution superior to that of scintillation cameras and have reasonable detection efficiency for low-energy photons from the decay of ^{125}I and ^{123}I . The synthetic collimator configuration allows for high-resolution tomographic reconstructions from data acquired with a single collimator–object angle. Therefore, we feel this system offers advantages over currently available small-animal imaging systems.

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