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Silicon detectors for combined MR-PET and MR-SPECT imaging

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Abstract

Silicon based devices can extend PET-MR and SPECT-MR imaging to applications, where their advantages in performance outweigh benefits of high statistical counts.

Silicon is in many ways an excellent detector material with numerous advantages, among others: excellent energy and spatial resolution, mature processing technology, large signal to noise ratio, relatively low price, availability, versatility and malleability. The signal in silicon is also immune to effects of magnetic field at the level normally used in MR devices. Tests in fields up to 7 T were performed in a study to determine effects of magnetic field on positron range in a silicon PET device. The curvature of positron tracks in direction perpendicular to the field's orientation shortens the distance between emission and annihilation point of the positron. The effect can be fully appreciated for a rotation of the sample for a fixed field direction, compressing range in all dimensions. A popular Ga-68 source was used showing a factor of 2 improvement in image noise compared to zero field operation. There was also a little increase in noise as the reconstructed resolution varied between 2.5 and 1.5 mm.

A speculative applications can be recognized in both emission modalities, SPECT and PET.

Compton camera is a subspecies of SPECT, where a silicon based scatter as a MR compatible part could inserted into the MR bore and the secondary detector could operate in less constrained environment away from the magnet. Introducing a Compton camera also relaxes requirements of the radiotracers used, extending the range of conceivable photon energies beyond 140.5 keV of the Tc-99m.

In PET, one could exploit the compressed sub-millimeter range of positrons in the magnetic field. To exploit the advantage, detectors with spatial resolution commensurate to the effect must be used with silicon being an excellent candidate. Measurements performed outside of the MR achieving spatial resolution below 1 mm are reported.

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Keywords

PET; silicon detectors

1. Introduction

High-resistivity silicon sensors as direct detectors of radiation offer a number of advantages in detection of radiation in nuclear medicine, among others:

- An excellent spatial resolution, with demonstrated resolution of $1\ \mu\text{m}$ [1] achieved with double sided strip detectors. The requirements are not nearly as harsh in nuclear medicine, where resolutions around 1 mm seem optimal.
- An excellent energy resolution of 1–2 keV FWHM in the range of signals between 59.5 and 511 keV [2].
- Mature processing and design technology, offering relatively cheap, highly configurable, sturdy and compact detectors.

Their most serious drawbacks are:

- A relatively low stopping power, with attenuation coefficient varying between 0.02 and $0.03\ \text{mm}^{-1}$ in the range of photon energies (140.5 to 511 keV) encountered in tracking of principal radionuclides. The low stopping power can be compensated by stacking of multiple sensors and proper multiplexing of the readout channels.
- Relatively poor timing performance related to long collection times. The problem can be circumvented with proper operating conditions, higher reverse bias applied over the sensor to speed up the charge collection, and modified sensor geometry, using interconnected thinner sensors to shorten the required drifting length. Both approaches combined yield devices that could be operated to activities of 50–100 MBq in and around the field of view [3] matching requirements posed by most applications in clinical imaging.

Nevertheless, silicon detectors could be used in targeted applications where one could exploit their advantages while compensating for their drawbacks. Two specific arrangements are presented, a Compton camera with silicon detectors [4] and a dual PET ring [5] with inner detector composed of silicon sensors.

Silicon detectors can be operated in magnetic fields without performance degradation [6], especially if the electric field in the device is parallel to the external magnetic field. For other relative orientations, the severe Lorentz angle [7] for electrons (55° for 8 T field) is partially compensated by signal of the holes where deflection is much smaller (14° at 8 T), pad geometry with a pad side slightly larger than the sensor thickness, and offline correction techniques. A silicon based device was built to study effects of magnetic field on imaging properties, more specifically the fact that the range of positrons is reduced in the direction perpendicular to the direction of the field due to circular motion of the positrons. A brief summary of the study will be presented.

2. Compton camera

Compton camera is a principle that can profit from usage of silicon detector. In a Compton camera, emitters of single photons are tracked based on Compton kinematics. Usually, the camera is composed of two detectors: a scatterer where initial photons interact through Compton scattering, generating Compton electrons to be analyzed by the sensor, and an absorber to capture the scattered photon. Impact positions in both sensors yield the scattered

photon track, and the energy E_e of the Compton electron is related to the scattering angle θ through Compton kinematics:

$$\sin^2 \frac{\theta}{2} = \frac{E_e}{E_\gamma - E_e} \frac{m_e c^2}{E_\gamma} \quad (1)$$

where E_γ is the initial energy of the photon, m_e is the electron mass and c the velocity of light in vacuum. At low energies, the principal component of the spatial resolution of the source position will come from limited angular resolution of the Compton collimation due to limited energy resolution of the sensor, ΔE_e :

$$\Delta \theta \propto \frac{\Delta E_e}{E_\gamma^2} \quad (2)$$

with rapid improvement with growing E_γ .

The principle was tested with a prototype that combined a silicon pad detectors with a cell size of $1.4 \times 1.4 \times 1 \text{ mm}^3$ as a scatterer coupled to low-noise readout electronics with a plain gamma camera with collimators removed and a resolution of 5–6 mm FWHM as an absorber [4]. The distance between the source and the scatterer was approximately 10 cm while the absorber was a further 20 cm away. Point sources of nuclides used in nuclear medicine ($^{99\text{m}}\text{Tc}$, ^{131}I , ^{22}Na) were placed on a rotary table and rotated in front of a slit collimator that compressed the imaging field to a single slice that contained the 1 mm thick silicon sensor. Images were reconstructed using a filtered back-projection algorithm. The resolutions obtained are shown in Figure 1 and compared to simulations.

At low energy, the Doppler broadening has a significant impact on the resolution. Modeling of the broadening has only recently been introduced to modeling software package EGS5 [8]. In the reference, the authors allow for certain discrepancies in treatment of momentum distribution of the bound electrons: firstly atomic rather than crystalline distributions are used and secondly, approximation rather than tabulated values were used in their estimation. Although observable in Figure 1, the effect was not further explored as the obtained results outline the predicted improvement in performance with growing E_γ .

A similar device could be used in magnetic fields allowing only the MR compatible part, the scatterer, to be inserted into the MR bore, while the absorber could be placed into a more relaxed environment further away from the magnet.

3. Dual ring PET

In a dual ring PET [5, 9] a silicon ring is used as insert to a standard PET detector ring. Events when one or both photons interact in the silicon detector exhibit excellent spatial resolution, contributing to improved image quality when combined with data obtained from the standard scanner. Calculations show that only a small fraction of events with a high spatial resolution are required making silicon a possible choice of the inner ring detector material.

A demonstrator of the dual ring geometry was constructed at the University of Michigan [5]. Concentrating only on the inner ring, detectors are placed as indicated in Figure 2, with two 1 mm thick silicon detectors placed behind a slit collimator that compresses the imaging field to a single, 1 mm thick slice through the imaging object, which is rotated for a full angular view. The imaging object is a Derenzo rod phantom with rod diameters ranging between 1.2 and 4.8 mm, filled with ^{18}F -FDG with peak activity of 185 MBq. Multiple 5

hour sessions were performed to accumulate 8.7 million counts used in MLEM reconstruction shown in Figure 3, indicative of a full ring performance. Resolution well below 1.2 mm can be recognized from the images.

Data was also collected in the volumetric PET mode, where the collimator was removed and the detectors were positioned perpendicular to the predominant direction of the annihilation photons. The same resolution was achieved when point sources of ^{22}Na were imaged.

4. PET ring in a magnetic field

A similar arrangement as in Figure 2 was placed in a 7 T MR magnet at the Ohio State University [6]. There was no degradation in performance of the sensors. The device was used to monitor changes in the range of positrons in a magnetic field. A ^{68}Ga source with maximum positron energy of 1.899 MeV was imaged both in and outside of the magnet. The resolution changed between from 2.3 to 1.6 mm FWHM when a magnetic field with a strength of 7 T was applied, allowing a separation of a pair of sources placed 3.6 mm apart in the magnetic field, which were inseparable in the 0 T field. The compressed positron range also shows an adverse effect on artifacts stemming from annihilations of positrons emitted by sources located off the plane of the selected object slice. Without the field, the off-plane sources contribute a relatively flat background which is indistinguishable from noise. In the field, the annihilations are sharply concentrated around the projection of the source on the selected slice, giving sharp artifacts in the reconstructed image. To remove the artifacts, the object must be imaged in multiple field orientations, posing additional requirements on a device to exploit the compressed range of the positrons in a magnetic field.

5. Conclusion

Benefits of silicon detectors can be exploited in targeted applications in nuclear medicine. Operation of silicon detectors is not perturbed by a presence of a magnetic field, which makes them feasible for targeted MR-SPECT or MR-PET applications as well. The high spatial resolution of silicon detectors is sufficient to monitor changes in positron range in magnetic field. To exploit the smaller positron range, the object must be rotated within the field during imaging.

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Figure 1.

Spatial resolution of the reconstructed point source measured with a Compton camera prototype. Horizontal axis indicates the energy of the photons used in reconstruction for source indicated next to the measurement points. Vertical axis shows the reconstructed resolution. Curves for measurement and simulation are shown.

Figure 2.

Geometry of silicon detectors and collimators used in dual ring prototype and in the device to measure the positron range in a magnetic field. The magnetic field, if present, was parallel to the axis of rotation of the object.

Figure 3.

Reconstruction of a single slit of a Derenzo rod phantom filled with ^{18}F -FDG, with hole diameters ranging from 1.2 to 4.8 mm as measured in the setup indicated in Figure 2. A total of 8.6 million events contributed to the image.