PET and SPECT Detector Developments and Applications to Multimodal Systems

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Introduction to Multi-Model Imaging
# An Overview of Mouse Imaging Systems


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<th>Technique</th>
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<td>Magnetic resonance imaging (MRI)</td>
<td>10–100 µm</td>
<td>No limit</td>
<td>Min/hours</td>
<td>Gadolinium, dysprosium, iron oxide particles</td>
<td>$$$</td>
<td>phenotyping, physiological imaging and cell tracking</td>
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<tr>
<td>X-ray computed tomography (CT)</td>
<td>50 µm</td>
<td>No limit</td>
<td>Min</td>
<td>Iodine</td>
<td>$$</td>
<td>Lung, bone, tumour imaging</td>
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<tr>
<td>Ultrasound imaging</td>
<td>50 µm</td>
<td>mm</td>
<td>Min</td>
<td>Microbubbles</td>
<td>$$</td>
<td>Vascular and interventional imaging</td>
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<tr>
<td>Positron emission Tomography (PET)</td>
<td>0.8–2 mm</td>
<td>No limit</td>
<td>Min</td>
<td>18F, 11C, 15O</td>
<td>$$$</td>
<td>Imaging metabolism of molecules, such as glucose, thymidine …</td>
</tr>
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<td>Single photon emission tomography (SPECT)</td>
<td>0.1-1 mm</td>
<td>No limit</td>
<td>Min</td>
<td>99mTc, 111In</td>
<td>$$</td>
<td>Imaging of probes such as antibodies, peptides …</td>
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<tr>
<td>Fluorescence reflectance imaging (FRI)</td>
<td>1–2 mm</td>
<td>&lt; 1 cm</td>
<td>Sec/min</td>
<td>Fluorescent proteinsNIR fluorochromes</td>
<td>$</td>
<td>Rapid screening of molecular events in surface-based tumours</td>
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<tr>
<td>Fluorescence-mediated tomography (FMT)</td>
<td>1–2 mm</td>
<td>&lt; 10 cm</td>
<td>Sec/min</td>
<td>NIR fluorochromes</td>
<td>$$</td>
<td>Quantitative imaging of targeted or ‘smart’ fluorochrome reporters in deep tumours</td>
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<tr>
<td>Bioluminescence imaging (BLI)</td>
<td>Several mm</td>
<td>cm</td>
<td>Min</td>
<td>Luciferin</td>
<td>$$</td>
<td>Gene expression, cell tracking</td>
</tr>
</tbody>
</table>

L. J. Meng, Harvard Medical School, April 2\textsuperscript{nd}, 2013
“Emission tomography and its place in the matrix of molecular imaging technologies”

Nuclear imaging offers tremendous sensitivity!!

Combining Nuclear Imaging with Other Imaging Modalities

FDG study: lung or ribs?

I-124 study: bone or soft tissue?

Two images is better than one!
Fused SPEM/CT image of a mouse’s brain. Two groups (5μL and 0.3μL) of radiolabeled T cells are visible in the brain. Meng et al, NIM, 2009.

99mTc-Labeled RGD Dimers and Tetramer in Tumor-Bearing Mice, Image courtesy Shuang Liu
Why Combine PET/SPECT with MRI

- Large variety of PET/SPECT tracers
- Sensitivity of PET/SPECT is in the picomolar range
- MR delivers exquisite soft tissue contrast, and functional imaging capabilities (spectroscopy, fMRI)
- No additional radiation dose (from CT)

Simultaneous imaging of PET and MRI
- Reduced total acquisition time
- Image multiple dynamic processes
- Use MRI to correct for motion in PET data

Martin S. Judenhofer, Seminars in Nuclear Medicine

Refresher Short Course RC4, MIC 2014
Why Combine PET/SPECT with MRI

FIGURE 1. First simultaneous PET/MRI study in 66-y-old healthy volunteer. MRI sequences included T2-weighted turbo spin echo, echo planar, time-of-flight MR angiography, and MR spectroscopy. PET image displayed was reconstructed from 20-min emission data recorded at steady state after injection of 370 MBq of 18F-FDG. Data were acquired on BrainPET prototype (Siemens). MPRAGE 5 magnetization-prepared rapid gradient echo; MRS 5 MR scintigraphy; TSE 5 turbo spin echo. (Ciprian Catana et al, Journal of Nuclear Medicine, 2012)
FIGURE 5. Simultaneous PET/MRI study in Alzheimer disease patient. (Top) Axial 18F-FDG PET, high-resolution MRI, and fusion image. Areas with reduced metabolism (green) representing impaired neuronal function are visible in left temporoparietal cortex. (Middle and bottom) Surface projections of cerebral metabolism and of z score images (comparison with controls). Data were acquired on Biograph mMR scanner. (Ciprian Catana et al, Journal of Nuclear Medicine, 2012)
Diagnostic Imaging with EM Radiation
Major Technical Challenges for Developing Combined Nuclear/MR Imaging Systems
Electromagnetic Interferences Between Nuclear and MR Data Acquisition Systems

Photomultiplier Tube (PMT)

http://nl.wikipedia.org/wiki/Fotomultiplicator

FIGURE 11 Depiction of the cross-wire anode PSPMT and its signal formation process. For simplicity, the charge avalanche created from only a single photoelectron is shown. A 10⁷ photoelectron amplification factor is typical. Setting the cathode at ground and biasing the anode at eHV through a coupling capacitor is also possible.


FIGURE 2 Scheme of dynode configurations: (a) venetian blind dynodes, (b) box dynode structure, (c) linear focusing dynodes, (d) circular cage dynodes, (e) mesh dynodes, and (f) foil dynodes. (From Photonics, 1994, Photomultiplier Tubes: Principles and Applications.)
Magnetic Field Induced Distortion

Measured (upper) and simulated (lower) energy responses of several adjacent pixels

(Above) Measured MR-induced projection shifting

Typical small-pixel CZT detector for use in MR scanners
Electromagnetic Interference Between Nuclear and MR Data Acquisition Systems

The drifting velocity of an electron under the electric field \( E \) and the magnetic field \( B \) is given by (Castoldi et al., NIM, 1997)

\[
\begin{bmatrix}
v_x \\
v_y \\
v_z
\end{bmatrix} = -\mu_{\text{red}} \times \begin{bmatrix}
1 + \mu_H^2 B_x^2 & \mu_H^2 B_x B_y - \mu_H B_z & \mu_H^2 B_x B_z + \mu_H B_y \\
\mu_H^2 B_x B_y + \mu_H B_z & 1 + \mu_H^2 B_y^2 & \mu_H^2 B_y B_z - \mu_H B_x \\
\mu_H^2 B_x B_z - \mu_H B_y & \mu_H^2 B_y B_z + \mu_H B_x & 1 + \mu_H^2 B_z^2
\end{bmatrix} \times \begin{bmatrix}
E_x \\
E_y \\
E_z
\end{bmatrix}
\]

where \( \mu \) is the charge carrier mobility and \( \mu_H \) is the Hall mobility. In the simplest form, where \( E=E_x \) and \( B=B_y \) then

\[
\begin{bmatrix}
v_x \\
v_y \\
v_z
\end{bmatrix} = -\frac{\mu}{1 + \mu_H^2 B^2} \times \begin{bmatrix}
E_x \\
0 \\
-\mu_H B_y E_x
\end{bmatrix}.
\]

The Lorenz angle \( \theta \) is given by

\[
\tan(\theta) = \mu_H B
\]
Interference Between Nuclear and MR Data Acquisition Systems

- Traditional PET detectors, based on PMTs, are sensitive to magnetic fields
- Magnetic field inhomogeneity induced by magnetic material in detector assembly and readout electronics
- MRI uses high frequency and high power RF which can interfere with the operation of PET detectors

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Major Challenges for Developing MR-Compatible Gamma Ray Detector

• **MR-compatibility** – minimized interferences between nuclear and MR imaging systems.

• **Compact foot-print** for better integration with other imaging modalities.

• **A reasonable balance** between spatial, energy and timing resolutions, detection efficiency and readout speed.

• **Cost-effectiveness and reduced complexity.**
Evolution of Scintillation Detectors
for (Combined MR-)PET and SPECT Applications
First approach to acquire PET data inside an MRI scanner (1997)

If the scintillation light is transported away from the high magnetic field, PMT detectors may be used.

Light loss due to fiber coupling is significant and degrades PET performance

MR field of view is free from metallic materials which preserves good MR performance

Optical Fiber Coupled Scintillator-PMT PET/MRI Detectors

- 2-layer depth of interaction PET detectors
- Full ring of 12 detectors
- Readout with position sensitive PMTs
- Good spatial resolution
- 0.3T MRI with hole in yoke used for MR imaging

Prof. S. Yamamoto, Kobe City College, Osaka University, Japan – Fibers with PSPMTs

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Optical Fiber Coupled Scintillator-PMT PET/MRI Detectors

University of Cambridge

- Split magnet MRI system
- Scintillator and fiber are in between magnet halves
- Read out with PMTs based on commercial small animal PET
- Special MRI coil design required

Problems of the fiber coupled PET detectors:
- Attenuation of scintillation light degrades both energy and timing resolution,
- limited physical design options, and
- Complexity.
(Gas) Multiplication Process in Solid-State Photon Sensors??

**Figure 6.3** Basic elements of a proportional counter. The outer cathode must also provide a vacuum-tight enclosure for the fill gas. The output pulse is developed across the load resistance $R_L$.  

**Figure 9.3.** Curve of pulse height versus voltage across a gas-filled pulse counter, illustrating the ionization chamber, proportional, and Geiger regions.

**Figure 7.1** The mechanism by which additional avalanches are triggered in a Geiger discharge.

Photodiode Detectors

Figure 1-1 Photodiode cross section

Typical reversed-biased photodiode detectors
- No internal gain
- Very high QE for red light (up to 90%)
- Limited timing resolution

Hamamatsu photodiode technical information
Avalanche Photodiode (APD)

Avalanche Photo Diodes (APD), proof to be suitable detectors for PET/MRI

- Could be operated inside strong magnetic field
- Added internal gain
- Fast timing property
- Gain critically depends on bias voltage
- Temperature stability could be an issue

http://learn.hamamatsu.com/articles/avalanche.html

Next Generation PET/MRI Systems

University of Tuebingen, Germany

- Larger FOV transaxial 7.2 cm/ axial 7cm
- LSO Blocks: 15x15 crystals (1.6x1.6x10mm³)
- 16 cassettes each 3 blocks
- Higher Sensitivity: 4-5% (estimated)

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Next Generation PET/MRI Systems

University of Davis, California, USA

- Located inside 7 Tesla MRI
- Use position sensitive APDs (14x14mm$^2$)
- 4 rings with 24 detector blocks (each),
- based on LSO arrays (1.2x1.2x14mm$^3$) and PSAPDs
- 60 mm axial FOV (whole body mouse)
- ~ 3-4% sensitivity (simulation)
- 1.3mm spatial resolution
Silicon Photomultipliers

Silicon Photo Multipliers (SiPM, Geigermode-APD)
a novel MR compatible PET detector

- G-APD 6.5 x 7 mm$^2$
- 3 x 3 mm$^2$, 60x60 cells
- Each cell 50 x 50 µm$^2$

- Each cell is operated above breakdown
- Output signal is sum of all cells
- Each cell provides maximum-gain signal on photon interaction

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Silicon Multiplier (SiPM) Detectors

FIGURE 1. (a) Simplified structure of a SiPM composed of G-APD cells. The G-APDs are joined together on a common substrate and are electrically decoupled. The outputs of the cells are connected to an Al grid used for the readout of the output signals. Each cell has a quenching resistor in series. (b) Each cell (G-APD) is a p–n junction with a very thin depletion layer between p+ and n+ layers. Drawings courtesy of Julien Bec, UC Davis.


FIGURE 2. (a) Simplified electric structure of a SiPM composed of several G-APDs in series with a quenching resistor. (b) Equivalent circuit of a single cell when the device is on (a bias voltage Vbias is applied) and is detecting photons. The capacitor CCell initially charged at Vbias discharges through RCell dropping the bias voltage to Vbreakdown. The avalanche process is quenched via the quenching resistor and then the device is recharged.
Silicon Multiplier (SiPM) Detectors

FIGURE. (a) Resistive network for a 4 x 4 SiPM array, (b) Crystal map acquired with a $^{22}$Na source irradiating a 4x4 array of 1.5 x 1.5 x 20 mm$^3$ LSO scintillator crystals coupled to the SiPM array.

FIGURE 3. Pulse height spectra from a 131 mm2 Hamamatsu MPPC S10362-11-025C acquired for three different light intensities (the red and blue curves correspond to the lowest and strongest intensities, respectively) showing peaks corresponding to different numbers of photoelectrons generated.

Silicon Multiplier (SiPM) Detectors
Sogang University, Korea – GAPD based PET insert

- based on 4x4 GAPD arrays
- 3x3x10 mm³ LYSO scintillators
- 16 block detectors
- 70 mm inner diameter / 13 mm axial
- 2.8 mm spatial resolution
- charge signal transmission readout

Kang J, et al., A small animal PET based on GAPDs and charge signal transmission approach for hybrid PET-MR
JInst. 6-P08012, Aug 2011
Silicon Multiplier (SiPM) Detectors

“Digital” Silicon Photo Multiplier

- Each cell’s bias voltage and output signal can be controlled
- A trigger network and embedded time to digital (TDC) converter provide fast timing information
- Energy is estimated by counting the pulses of fired cells
- Cells with bad performance can be deactivated online

![SiPM Microphotograph](image)

~8100 cells

Conventional readout of SiPM for PET

![Diagram of conventional readout](image)

Readout of “digital” SiPM for PET

![Diagram of digital SiPM](image)

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Silicon Multiplier (SiPM) Detectors for PET Applications

HYPER Imaging Project, RWTH-University, Aachen

22 x 22 = 484 LYSO crystals, each 1.32 x 10 mm³
SiPM with 42 mm² in monolithic arrays of 2 x 2 SiPMs
2 x ASICs with 40(32) channels with individual EDC/TDC
Parallel FPGA data processing

Small animal PET insert build for 3T Human MRI

Slide Courtesy: Volkmar Schulz, RWTH

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**INSERT: INtegrated SPECT/MRI for Enhanced Stratification in Radio-chemo Therapy**

**SEVENTH FRAMEWORK PROGRAMME**

GA n. 305311  
Kickoff: 01/03/13  
Duration: 4 years

**INSERT members**
- Politecnico di Milano (Italy)
- Mediso Medical Imaging Systems (Hungary)
- Fondazione Bruno Kessler (Italy)
- Nuclearfields International BV (Netherlands)
- MRI.Tools GmbH (Germany)
- University College London (UK)
- Universita Vita-Salute San Raffaele (Italy)
- Universita Degli Studi di Milano (Italy)
- Cromed Research and Services ltd. (Hungary)
- CF Consulting srl. (Italy)

**Goal:** to provide improved personalized radio-chemo therapies for brain tumour (Glioma) patients using a specifically developed multi-modality imaging tool  
*Salvado et al. Proc IEEE NSS/MIC 2014; images courtesy UCL, Nuclear Fields*
Clinical system:

**MRI:**
- 3 T MRI
  (internal bore diameter ~60 cm)
- Customized RF coil

**SPECT:**
- Stationary system
- Multi Slit-Slat collimator
- 25 independent detection modules (FOV ~ 10x5 cm²)
- Foreseen radiotracers:
  - $^{99m}$Tc (140 keV)
  - $^{123}$I (159 keV)
  - $^{111}$In (171/254 keV)

*image courtesy Medisc*
Detection module:
• Monolithic slanted scintillator (CsI:Tl. Area~10x5 cm^2. Thickness 8 mm)
• Silicon PhotoMultiplier matrix
• ASIC readout and Data Acquisition System

Expected performance from Monte Carlo simulations:
➢ Intrinsic spatial resolution: between 0.8 and 1.0 mm
➢ Energy resolution: between 11% and 15% (Tc-99m - 140 keV) depending on the dark count rate of the SiPM technology (100-500 kHz/mm^2).

Busca et al. **M08-5**, IEEE NSS/MIC 2014; image courtesy Polimi. Milan
Collimator design:

➢ Novel ‘internal’ Multi-Slit-Slat (MSS)
  – Slit-slat: combines pinhole collimation in the transaxial and parallel-hole collimation in the axial dimension
  – Array of short slits placed internal to slats
    – Improves radial sampling
    – Optimal use of limited space

Salvado et al. M12-2, IEEE NSS/MIC 2014; images courtesy UCL, Nuclear Fields
Simulated results:

Reconstructed Images
• Simulations of digital phantoms (uniform, Derenzo, Defrise) with attenuation included, but not scatter
• Projection: based on angular blurring
  [Bousse et al. Fully 3D Recon Meeting 2013]
• Reconstruction: MLEM based on the same projector

slightly better transaxial resolution

~40% better sensitivity and uniform axial coverage
### Pros and Cons of PMT, APD and SiPM Detectors

<table>
<thead>
<tr>
<th></th>
<th>PMT</th>
<th>APD</th>
<th>Si-PM</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Amplification</strong></td>
<td>$10^6$</td>
<td>$10^2$</td>
<td>$10^6$</td>
</tr>
<tr>
<td><strong>Magnetic Field</strong></td>
<td>Sensitive</td>
<td>Not sensitive</td>
<td>Not sensitive</td>
</tr>
<tr>
<td><strong>Bias Voltage</strong></td>
<td>1000V</td>
<td>350-2000V</td>
<td>20-70V</td>
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<tr>
<td><strong>Signal / Noise Ratio</strong></td>
<td>High</td>
<td>Relatively Low</td>
<td>High</td>
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<tr>
<td><strong>Dynamic Range</strong></td>
<td>High</td>
<td>High</td>
<td>Small</td>
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<tr>
<td><strong>Timing Properties</strong></td>
<td>&lt; a few hundred ps – a few ns</td>
<td></td>
<td></td>
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<tr>
<td><strong>Electronic Readout</strong></td>
<td>Voltage Amplifier</td>
<td>Charge sensitive pre-amplifier</td>
<td>Voltage Amplifier, digital read out</td>
</tr>
</tbody>
</table>

**Towards:**
- Large internal gain, high signal-to-noise ratio
- Simplified high-speed readout electronics
- Solid-state detectors, very compact and potentially low cost
- Immune to strong magnetic field induced distortions
Semiconductor Detectors for Gamma Ray Imaging – Basic Principle

http://www.mdpi.com/1424-8220/13/2/2447/htm
### Comparison of Semiconductor Materials

<table>
<thead>
<tr>
<th>Detector Material</th>
<th>Z</th>
<th>( E_g ) (eV)</th>
<th>( E_{\text{pair}} ) (eV/ehp)</th>
<th>Density (g/cm³)</th>
<th>Resitivity @ 300K (( \Omega \cdot \text{cm} ))</th>
<th>( \mu \tau_{e/h} ) (cm²/V)</th>
<th>Knoop Hardness</th>
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<tr>
<td>Ge</td>
<td>32</td>
<td>0.66</td>
<td>2.9</td>
<td>5.33</td>
<td>50</td>
<td>&gt;1/~1</td>
<td></td>
</tr>
<tr>
<td>Si</td>
<td>14</td>
<td>1.12</td>
<td>3.6</td>
<td>2.33</td>
<td>( \sim 10^4 )</td>
<td>&gt;1/~1</td>
<td>1150</td>
</tr>
<tr>
<td>CdTe</td>
<td>48/52</td>
<td>1.4</td>
<td>4.4</td>
<td>6.2</td>
<td>( 10^9 )</td>
<td>( 10^{-3} / 10^{-4} )</td>
<td>45</td>
</tr>
<tr>
<td>CdZnTe</td>
<td>48/30/52</td>
<td>1.6</td>
<td>4.7</td>
<td>( \sim 6.2 )</td>
<td>( 10^{11} )</td>
<td>( 10^{-3} / 10^{-6} )</td>
<td></td>
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<tr>
<td>HgI₂</td>
<td>80/53</td>
<td>2.1</td>
<td>4.2</td>
<td>6.4</td>
<td>( 10^{13} )</td>
<td>( 10^{-4} / 10^{-6} )</td>
<td>&lt;10</td>
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<tr>
<td>GaAs</td>
<td>31/33</td>
<td>1.4</td>
<td>4.3</td>
<td>5.32</td>
<td>( 10^8 )</td>
<td>( 10^{-5} / 10^{-6} )</td>
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<tr>
<td>Diamond</td>
<td>6</td>
<td>5</td>
<td>13</td>
<td>3.51</td>
<td>( &gt;10^{13} )</td>
<td>( 10^{-5} / 10^{-5} )</td>
<td>10000</td>
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<tr>
<td>TlBr</td>
<td>81/35</td>
<td>2.7</td>
<td>5.9</td>
<td>7.56</td>
<td>( 10^{11} )</td>
<td>( 10^{-5} / 10^{-6} )</td>
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<tr>
<td>InP</td>
<td>49/15</td>
<td>1.4</td>
<td>4.2</td>
<td>4.78</td>
<td>( 10^7 )</td>
<td>( 10^{-5} / 10^{-5} )</td>
<td></td>
</tr>
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</table>


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Semiconductor Detectors for Gamma Ray Imaging

Compared to scintillation detectors, semiconductor detectors generally require

...... less energy (several eV versus a few hundred eV) to create an information carrier (electron/hole pair).

Therefore, direct-detection semiconductor detectors could potentially offer improved SNR, and therefore better spatial, energy resolution.
However, there are several challenges for using semiconductor detectors

- Poor charge collection efficiency – especially due to the poor mobility-lifetime product for the holes.
- Intrinsically slow signal generation process – poor timing resolution, and relatively low count-rate capability
- No internal gain, the need for low-noise amplifier circuitry
- Generally low detection efficiency per unit detector volume
- More expensive than scintillation detectors.
Room-Temperature Semiconductor Detectors for Gamma Ray Imaging – Basic Operating Principle

(Right) Illustration of a conventional detector using planar electrodes. Top: The schematic of a semiconductor g-ray detector using conventional planar electrodes. Middle: The weighting potential of the anode, and signal components from the movements of electrons and holes. Bottom: The expected energy spectra with fixed energy deposition $E_g$ when the g-ray interaction depth is distributed uniformly between the cathode and the anode.
Single Polarity Charge Sensing based on the Small Pixel Effect to Alleviate the Effect of Poor Hole Collection Efficiency


Fig. 5. Illustration of 3-D position sensitive semiconductor devices. (a) Pixelated anode array used by He et al. [35]. (b) Schematic of anode array suggested by Hamel et al. [39]. (c) Illustration of a device suggested by Luke et al. [41].

**Single Polarity Charge Sensing based on the Small Pixel Effect to Alleviate the Effect of Poor Hole Collection Efficiency**

Signal from anode pixel $\Rightarrow$ No. of electrons collected by the pixel (N) and its lateral position $(x, y)$.

$C/A$ $\Rightarrow$ Interaction depth $(z)$.  

$N, x, y, z$ $\Rightarrow$ Energy deposition $E_0$ and interaction location.
Timing Readout to Provide 3-D Position Information
Developed by Zhong He, Univ. of Michigan

- For multiple interactions, C/A ratio is no longer sufficient for determining interaction sites.
- Extra information is provided by drifting times for each electron cloud.
- The accuracy of determining interaction depth is <0.5mm
The Use of Transient Signal to Provide Spatial Resolution Smaller Than the Actual Pixel Size

Yuefeng Zhu, Stephen E. Anderson, and Zhong He

IEEE TRANSACTIONS ON NUCLEAR SCIENCE, VOL. 58, NO. 3, JUNE 2011

Fig. 1. An illustration of the pixelated CdZnTe detector used in this study.

Fig. 2. Signal induction for a collecting pixel (pixel 22) and its 8 neighbors. The responses correspond to a single simulated 662-keV point electron cloud collected by the center pixel. The transient signals of the neighbor pixels are shown for two events: one happened underneath the center (thick line) and the other near the edge (dashed line) of the collecting pixel and close to pixel 21. They are both located in the middle depth of the detector.
The Use of Transient Signal to Provide Spatial Resolution Smaller Than the Actual Pixel Size

Yuefeng Zhu, Stephen E. Anderson, and Zhong He
IEEE TRANSACTIONS ON NUCLEAR SCIENCE, VOL. 58, NO. 3, JUNE 2011

Fig. 6. The collimator design for experimentally measuring sub-pixel position resolution.

Fig. 8. The simulated distribution of the electron cloud centroid from a 662-keV gamma-ray source using a collimator. Distribution width is due to the collimator and electron cloud size.
Double-Sided Strip Readout to Reduce the Number of Readout Channels and Reduce the Complexity of Large Area Imaging Detectors

Henric Krawczynski, Washington University in St. Louis, Physics Department and McDonnell Center for the Space Sciences
Sharing the Readout Duty by Anode-Pixel Circuitry and Cathode Waveform Sampling to Simplify Small-Pixel CZT and CdTe Detectors

Pixel readout circuitry to provide:
- Pixel address
- Coarse timing info for synchronizing the cathode and anode readout operation
- Coarse energy information (if needed)

Waveform sampling circuitry to provide:
- Precise timing
- Accurate depth of interaction (DOI) information
- Energy information

Benefits (presented at IEEE RTSD 2011):
- Highly simplified pixel-circuitry, pixel address and triggering only.
- Improved timing and DOI resolution (for single interaction).
- Independent of anode configurations – allow the use of further reduced pixel sizes and therefore further improved spatial resolution.
Study of a high-resolution, 3D positioning cadmium zinc telluride detector for PET

Y. Gu et al., Physics in Medicine and Biology

Figure 4. Pictures of the CZT detector and flex circuits patterned with traces that match the pitch of the cross-strip electrode pattern. (a) CZT detector resting on flex circuits with the cathode side up, and (b) flex circuits coupled to detector electrodes.

Figure 5. CZT photon detector electrical and mechanical connections. (a) Detector’s high-voltage bias and electronic connections, and (b) detector’s mounting assembly.
Development of A MR-Compatible SPECT System
Ideas Ltd Norway, and Johns Hopkins University
Development of A MR-Compatible SPECT System
Ideas Ltd Norway, and Johns Hopkins University

Cadmium Zinc Telluride (CZT) and Application Specific Integrated Circuits (ASICs)

Digital Radiation Detector

Image courtesy Dirk Meier, IDEAS, Norway

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New Configurability of Imaging Systems

- 50 years planar NaI(Tl) Anger camera
- New: curved detector surface, enabled by flexible circuit boards
- Arcs can be formed for closer proximity, reduction in parallax

Image courtesy Dirk Meier, IDEAS, Norway

Refresher Short Course RC4, MIC 2014
Development of A MR-Compatible SPECT System
Ideas Ltd Norway, and Johns Hopkins University

32 CZT modules SPECT Prototype for 12-cm MRI bore

Image courtesy Dirk Meier, IDEAS, Norway
Refresher Short Course RC4, MIC 2014
Pixelated CdTe Detectors Developed for MRC-SPECT Applications

- Pixelated CdTe (above, left) and CZT (below, left) detector of 11mm × 22mm × 1 or 2 mm in size and having 32×64 350 μm × 350 μm pixels.
- ERPC detectors with 2 mm thick CdTe detectors will be used in the prototype system.
- Other pixel sizes – 515 um, 700 um read out with the same ASIC?
The emphasis of this development is trying to simplifying this circuitry as much as possible.
Miniaturization of the ERPC Detectors (2007-2012)

FPGA for controlling the readout sequence
Detector hybrids 1.1 cm × 2.2 cm
Wire-bonding to the readout PCB

The area within the red square represents a detection area of 1.1 cm × 2.3 cm
Non-magnetic Ultrahigh Resolution CZT/CdTe Detectors

A prototype MRI-Compatible ERPC detector

- CdTe or CZT detectors of 2-5 mm thickness.
- Supported on a ceramic substrate, also served as heat sink, cooled with compressed air.
- Low profile design.
- Non-magnetic construction.

Actual MRC-SPECT detector module

- 4.5 cm x 2.25 cm.
- 350 μm x 350 μm pixels.
- 2 mm thickness CdTe, or 5 mm CZT.
- Either frame-by-frame, on-chip counting, or photon-counting mode.
- Minimized ferromagnetic materials.
The MRC-SPECT System as a Standalone Desktop SPECT System and an Integrated SPECT/MRI System

The MRC-SPECT system inside a Siemens MR scanner.

The MRC-SPECT system as a standalone desktop high-performance SPECT system.
Thank You!

and Any Questions?