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sense and simplicity
State of the Art and Future Trends in Radiation Detection for Computed Tomography

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Global Research & Advanced Development
CT BU, PHILIPS Healthcare
Plan

1. Conventional, state of the art, and future CT detection system

2. A Dual-Layer detector for a Detector- Based Spectral CT, with some applications’ examples

3. Photon-Counting Spectral CT detectors, advantages, opportunities, and risks
A Little About CT

Present CT

1. Large Coverage Detector
2. Wedge Configuration
3. Fast Rotation (close to 0.25 sec/Rot.)
4. 2D Focal-Spot double sampling
5. Very Short angular sampling time (~100 µsec)
6. High-Rate and Power X-Ray Tubes
7. Sub-Millimeter isotropic resolution
8. Very good temporal resolution
9. A 3D and 4D imaging devise (Space-Time)
So How Does It Look Like in Reality?
### Modern CT Detectors Requirement

<table>
<thead>
<tr>
<th>Requirement</th>
<th>Specification</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frame Rate &amp; size</td>
<td>10,000 frames/sec</td>
</tr>
<tr>
<td></td>
<td>~120,000 pixels/frame</td>
</tr>
<tr>
<td>Detector Readout Mode</td>
<td>Current (energy) Integration</td>
</tr>
<tr>
<td>Electronic noise RMS</td>
<td>&lt; 3 pA</td>
</tr>
<tr>
<td>High light output scintillator</td>
<td>Ceramic Gd$_2$O$_2$S (GOS) (~40,000 photons/MeV)</td>
</tr>
<tr>
<td>Low Scintillator Afterglow</td>
<td>&lt;200 ppm at 3 m-sec</td>
</tr>
<tr>
<td></td>
<td>&lt;20 ppm at 500 m-sec</td>
</tr>
<tr>
<td>Expandable configuration</td>
<td>4-sides tile-able arrays</td>
</tr>
<tr>
<td>Spatial Resolution</td>
<td>~24 lp/cm (~0.210 mm)</td>
</tr>
<tr>
<td>Scattering rejection</td>
<td>SPR &lt; 5%; 2D Anti-Scatter-Grid</td>
</tr>
<tr>
<td>Maximum Crosstalk (optical &amp; elec)</td>
<td>~5%</td>
</tr>
<tr>
<td>Pixelated Si photodiode with excellent response</td>
<td>&gt; 0.35 A/Watt for $\lambda = 514$ nm</td>
</tr>
<tr>
<td>Dynamic Range</td>
<td>$\geq 2^{20}$</td>
</tr>
<tr>
<td>Scintillator Stopping power</td>
<td>&gt; 98% for 120 kVp spectrum</td>
</tr>
</tbody>
</table>

Signal Pixel $\propto$ Total X-Ray Energy absorbed in the scintillator within a sampling time.
Pixelated Large-Area Photodiode for CT Detector

Front-illuminated photodiode: anodes directly illuminated; limited expandability

Back illuminated “64 slices” photodiode “flip-chipped” to a substrate

Si Photodiode response
# Scintillator Arrays for CT, Material types

## CT Scintillators in Use, (Potential Use)

<table>
<thead>
<tr>
<th>Scintillator</th>
<th>Light-yield (# photons/M eV)</th>
<th>Form: Ceramic \ Single Cristal</th>
<th>Afterglow</th>
<th>Comment</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gd$_2$O$_2$S:Pr,Ce (GOS)</td>
<td>~40,000</td>
<td>Ceramic; semi translucent</td>
<td>low</td>
<td>Most vendors; some doping variations</td>
</tr>
<tr>
<td>GE Gemstone™ (a Lu based Garnet)</td>
<td>~40,000</td>
<td>Ceramic; translucent</td>
<td>low</td>
<td></td>
</tr>
<tr>
<td>Garnet type of (Lu,Gd,Y,Tb)$_3$(Ga,Al)$<em>5$O$</em>{12}$</td>
<td>40,000-45,000</td>
<td>Ceramic; translucent</td>
<td>low</td>
<td>Fast rise time</td>
</tr>
<tr>
<td>(Y,Gd)$_2$O$_3$:Eu (GE HiLight™)</td>
<td></td>
<td>Ceramic</td>
<td>high</td>
<td>not adequate for short Integration Periods</td>
</tr>
<tr>
<td>ZnSe:Te (low stopping power)</td>
<td>~65,000</td>
<td>Single Cristal; semi-translucent</td>
<td>low</td>
<td>In Philips Dual-Layer CT prototype</td>
</tr>
</tbody>
</table>

$$\text{Light \_Yield \_Limit} \approx \frac{10^6}{2.5 \cdot E_g} \text{photons / MeV}$$

**NOTE:** Light output is smaller than Light-Yield because of Internal absorption & Reflections
Pixelated Scintillators, and CT DAS Assembly Modes

Philips Brilliance-64 Module W/O and with anti-scatter grid

PHILIPS iCT-256, Tiled Config., 2D anti-scatter grid

2D-Anti-Scatter grid

Tiled Module

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Why Anti Scatter Grids in CT Detectors?

1. Detector-Pixel signal at each sampling is assumed to represent a line integral of the form:
   \[ P(\phi, \theta) = \ln(I_0) - \ln(I) = \int_{\xi_{in}}^{\xi_{out}} \mu(\xi)d\xi \]
   that assumes X-Ray pure-transmission only.

2. Allowing scattered radiation \(\Rightarrow\) Cupping and blackened streaks artifacts between scattering centers:

   1D VS. 2D Anti-Scatter Grid

   SW Scatter-artifacts Correction effect!
   Note the significant noise increase.

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Estimated upper limits of MTF and various CT parameters contribution to it in standard resolution mode (Philips iCT-256), assuming Sinc function response:

\[
DQE = \frac{SNR^2_{out}}{SNR^2_{in}}, \text{ or measured: } DQE(f) \approx \frac{S^2 MTF^2(f)}{NPS(f) \cdot \phi}
\]

(S=detector signal; \(\phi\) = # of photons per area unit; NPS=Noise Power Spectrum)

A measured single pixel MTF and DQE (PHILIPS tile detector), (including Swank Noise and Crosstalk (R Luhta et al. SPIE-2006))

Overall Detection Efficiency (DE) has to include the Geometrical Detection Efficiency (GDE), determined by the fraction of active area to total pixel area in the pixelated scintillator (~73%).

\[
DE = GDE \cdot DQE
\]
Measuring Detection-Pixel crosstalk and deviation from pure square wave response due to crosstalk.  
(R. Luhta et al, SPIE-2006)

Correcting for Pixel Crosstalk, and Scintillator Afterglow through deconvolution method (Philips Brilliance 16), in an Ultra-High Resolution Mode (slice plane)

(See R. Carmi et al. Nuclear Science Symposium conference record 2004, 5:2765-2768.)
Photodiode’s current as a function of absorption path (water), and Tube current

- An exaggerated RMS noise of 5 pA has been taken in account, just for demonstration
- Today’s readout ASICS which are assembled attached to the photodiodes have RMS noise < 3 pA
- In most read-out electronics, both Sigma-Delta, or Current-to-Frequency, the RMS noise increases slightly for shorter integration periods
Low Dose Artifacts (rings), introduced by Electronic Noise and Non-Linear Effects

1. 50-cm water phantom
2. Old version Electronics Philips BR-64 (non-tiled)
3. Scanned at 0.4 sec/rot., with 70 mAs
4. Measured RMS noise ~5pA
5. Artifacts caused mainly by electronics non-linear effects and offset stability during the scan
Dual-Energy Spectral CT Based on A Dual-Layer Detector

A 0.125-mm side-looking photodiode array shielded by a 0.5 mm Tungsten layer

0.030 mm optical glue

0.080 mm reflecting paint

X-Rays Coming from top

Bottom Scintillator:
2-mm GOS

Top Scintillator, (e.g. ZnSe) 1.0 mm

~50%

→ Low Energy Raw data → E1 image

~50%

→ High Energy Raw data → E2 image

= Weighted combined Raw data → CT image

Stacking Dual-Layer 1-D arrays into a single standard detection tile

A 16 X 16 pixels Dual-Layer Detector Tile for A Dual-Energy Spectral CT
A phantom with different concentrations of Calcium and Iodine contrast agent.

\[
\mu_{Z,E} \approx A \cdot \frac{Z}{E^{0.2}} \cdot \rho + B \cdot \frac{Z^3}{E^3} \cdot \rho
\]

Conventional CT Image: Soft Tissue, Calcium & Bones, Fat

Spectral CT Image, Dual-E: Soft Tissue, Iodine-tagged Blood, Calcium, Fat

Spectral Analysis Map: HU of E1, HU of E2

Water (E_low & E_high = 0 HU)
## Dual-Energy Technologies – Various Vendors

<table>
<thead>
<tr>
<th>Technology Path</th>
<th>Siemens</th>
<th>GE</th>
<th>Philips</th>
</tr>
</thead>
<tbody>
<tr>
<td>Two-tube Two-Detector configuration</td>
<td>Two-tube Two-Detector configuration</td>
<td>One tube with fast kV switching</td>
<td>One tube, detectors with simultaneous high and low energy discrimination</td>
</tr>
<tr>
<td>Full FOV</td>
<td>✗ Limited (~35 cm)</td>
<td>✓ Full (50 cm)</td>
<td>✓ Full (50 cm)</td>
</tr>
<tr>
<td>High/Low Energy Separation</td>
<td>✓ +</td>
<td>✓</td>
<td>✗</td>
</tr>
<tr>
<td>Projection Space Reconstruction</td>
<td>✗</td>
<td>✓</td>
<td>✓</td>
</tr>
<tr>
<td>Tube Current Modulation</td>
<td>✗</td>
<td>✗</td>
<td>✓</td>
</tr>
<tr>
<td>Low/High Energy Image Reconstruction</td>
<td>✓</td>
<td>✗</td>
<td>✓</td>
</tr>
<tr>
<td>Retrospective Dual-E Analysis, all protocols</td>
<td>✗</td>
<td>✗</td>
<td>✓</td>
</tr>
</tbody>
</table>
Dual-Layer Spectral CT – Main Clinical Applications

- Direct CTA
- Gout Diagnosis
- Prep-Less CT Colonography

- Virtual Non-Contrast
- Blood Flow Iodine Perfusion (PE)
- Lesion Uptake & Volume Assessment
- Plaque Characterization
- Cardiac CT
- Urinary Stone Characterization

- Salvaging Sub-Optimal CTAs
- Optimum CNR Imaging (↑Lesion visualization)
- Metal & Beam Hardening Reduction
- Effective Atomic Number
Dual-Layer Spectral CT, Clinical Experience

Monochromatic Imaging

Lesion Characterization

Material Decomposition

Material Decomposition (Color Display)
Dual-Layer Spectral CT, Viewer Screen Shot Example

MonoE Imaging

Spectral Attenuation

Optimum CNR

Effective Z

Selecting ROIs

Material

Reference
High energy photons hit a semiconductor and are directly converted to an analog signal. A fast ASIC translates individual signal pulses to counts at several energy bins.

~ 13,000 e-hole pairs per 60 keV photon

~ 1000 e-hole pairs per 60 keV photon in the photodiode

Integrated Energy
(Discrimination and counts)

Spectral Profile
(white light intensities)
Photon Counting Detector Direct Conversion VS Energy Integration

A Current-Integration CT Detector

Operation Principles – A CT Photon-Counting Detector

A Photon-Counting, Direct-Conversion, CT Detector

PhC Requires much smaller pixels to enable CT radiation rates
The Photon Counting Energy Weighting “Miracle”

\[ S_{ph.c.} \propto \sum_{i=1}^{N_{bins}} 1 \cdot k_i \quad S_{EI} \propto \sum_{i=1}^{N} E_i \cdot k_i \rightarrow \int_{0}^{\infty} E \cdot w(E) dE \]

\[ \sigma_{ph.c.} \propto \sum_{i=1}^{N_{bins}} 1^2 \cdot k_i \quad \sigma_{EI} \propto \sum_{i=1}^{N} E_i^2 \cdot k_i \rightarrow \int_{0}^{\infty} E^2 \cdot w(E) dE \]

1. \( Z_{eff} \) of Plexiglas is \(~20\% < \text{water}\
2. Energy Integrating over densities over \( Z \).
3. Photon Counting changes HU ratios, Contrast difference, and CNR
4. Opportunity to adapt energy weighting to specific applications
5. Potential increase in soft-tissue resolution.

Ami Altman, Ph.D., PHILIPS Healthcare

I – iodine
Gd – gadolinium
Ca – calcium (as CaCl)

H – high concentration
L – low concentration

C – solid carbon
A – air bubbles (in foam)

(Measured in a Rotating Table Photon Counting Mini CT, PHILIPS Healthcare, Haifa)

Mixtures or interfaces of more than 2 materials, suffers from spectral partial volume, distorting material decomposition and classification

2 spectral windows: 25-46, 85-125 keV

6 spectral windows: 25-46, 46-54, 54-65, 65-75, 75-85, 85-125 keV
So Where is The Catch, Why CT is Not Transforming to Photon-Counting Detector?

1. Best Candidates, CdTe and CdZnTe (CZT) are not easy to manufacture, to meet CT requirements

2. Detector Polarization (eliminating or distorting the electric field inside the detector, stopping any signal output), mainly caused by traps and impurities in the crystal lattice

3. Charge sharing between close neighboring pixels, causing low energy tail, that distorts measured energy spectrum, especially for a wideband spectrum like in CT

4. Te and Cd fluorescence K-escapes to neighbor pixels and from neighbor contributing to the low-energy tail

5. Pile-Up already at relatively low rate
Pile-Up Issue in Photon Counting

OCR curves for $\tau = 30$ nsec

Paralyzable: $OCR = ICR \times \exp(-ICR \times \tau)$

Non-Paralyzable: $OCR = \frac{ICR}{1 + ICR \times \tau}$

OCR=Output Count Rate
ICR=Input Count Rate

(K. Taguchi, SpecXray 2011 CERN)
DQE(0) in Photon Counting Detectors

\[ DQE(0) = \left( \frac{\partial \text{OCR}}{\partial \text{ICR}} \cdot \left( \frac{\text{ICR}}{\sqrt{\text{VAR}_{\text{OCR}}}} \right) \right)^2 \]

For Variance estimates see

Low-Energy Tail and its effect on Spectral CT Quality

CdTe arrays: Spectral response to Am241 source (59.5 keV)


CdTe arrays: Spectral Response to a 110 kVp X-Ray Tube
Pre-clinical spectral CT scanner platform

- Gantry with rotation speed up to 1/3 s per turn
- \(\mu\)-Focus X-ray tube
  - 40 kVp - 130 kVp
  - max. 65W
- CdTe-based Photon-Counting detector
  - 0.4 mm pixel pitch, 1024 pixel along X
  - 6 energy bins per pixel
- Magnification: 2 - 6
- Field-of-view: 6 cm - 23 cm
- Spatial Resolution: 100 \(\mu\)m - 250 \(\mu\)m
Enabling K-Edge Imaging and Targeted CT Imaging (PHILIPS – Mount Sinai Collaboration)
Washington University Collaboration

Imaging nano particles, loaded with Gadolinium & Bismuth, and targeted mainly to fibrin in soft plaque

Anatomical mouse image with iodine concentration overlay

Human carotid samples with targeted contrast agent

Artificial calcification inside of a stent.

Left: integrated CT image, Right: photo- and Gd image

Hepatitis steatosis (fatty liver) phantom measurements
imaging results
From collaboration with Eindhoven

- iodinde-based bloodpool agent
- mouse images

Tail cross section

<table>
<thead>
<tr>
<th>organ</th>
<th>view</th>
<th>$\rho_f$ [$\mu g/mm^3$]</th>
</tr>
</thead>
<tbody>
<tr>
<td>liver</td>
<td>c</td>
<td>4.6 ± 0.2</td>
</tr>
<tr>
<td>spleen</td>
<td>c</td>
<td>4.0 ± 0.3</td>
</tr>
<tr>
<td>kidney</td>
<td>c</td>
<td>1.9 ± 0.2</td>
</tr>
<tr>
<td>heart</td>
<td>s</td>
<td>2.9 ± 0.2</td>
</tr>
<tr>
<td>liver</td>
<td>s</td>
<td>4.5 ± 0.2</td>
</tr>
</tbody>
</table>
Applications demand high delivered payloads of heavy metals targeted to intraluminal thrombus, but excluded from intraplaque fibrin.

Nanoparticles home specifically to intra-lumenal fibrin not intramural fibrin from healed hemorrhage or rupture (in vitro CEA specimens from Patient)

Pan, Roessl, Thran,…Proksa, et. al. Angew Chem Int Ed. 2010, 9635-39
Investigating K-Edge Imaging at WashU

Imaging of new Ytterbium Nanobeacons

Pan, Schirra, Roessl, Thran, … , Proksa et al., ACS Nano. 2012 (in press, e-pub available)
Consequences

1. Innovative detectors technologies has lead, and will lead future CT evolution and revolution towards quantitative imaging in more than just anatomy and morphology.

2. Detector-based spectral CT enables the benefit of material\tissue analysis retrospectively for all conventional protocols, without the need to decide on a specific Dual-Energy protocol.

3. Photon-Counting detectors for spectral CT imaging is a promising technology, that offers also much better spatial resolution, no electronic noise, better CNR, capability to further reduce dose, and a more accurate material and tissue representation and decomposition.

4. Photon-counting CT may open an opportunity to use targeted contrast agent, thus become more quantitative, enabling more personalized diagnostics.

5. We are not yet there. There are several significant technology challenges, in detector performance and pile up issues, to overcome.