Advances in Photon Counting Detectors in X-ray Imaging:  
*Photon Counting Detectors for Mammography*

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Disclosure statement:  
- Founder Sectra Mamea  
- Shareholder Sectra

Work in progress - FDA approval pending for all devices described in this presentation

To Count Photons – What does it mean?  

Result:  
- Number of X-rays  
- Pulse height gives the color for each X-ray  
- No electronic noise (regardless of pixel size and dose)
Integrating Current: Today’s praxis in X-ray Imaging

**X-ray photons**

Energy vs. Time

Result:
- Area of
- Indirect estimate of number of photons

Mammography

- 100 M women examined every year
- Mortality in breast cancer reduced 30-50%
- Glandular tissue sensitive to radiation
- High requirements on spatial and contrast resolution
- Low kVp
- Relatively low count rate
- Object relatively still

Photon Counting Mammography Today

Sectra – Silicon detectors
- 200 systems in 15 countries

XCounter – Gas detectors
Research and development

Both are scanned-slit

In the basement Sectra mammography started 9 years ago..
First model 2003 (D20) “heavy and large”

Current model 2007 (L30) “light and slim”
Scanned Slit – Efficient Scatter Rejection

Geometry for photon counting (diamonds) increases system DQE with 40% compared to typical grid

Silicon for Photon Counting

Detector Response to High Energy X-rays

Cross sections for interaction

Silicon for Photon Counting

Energy

Intensity

Compton edge

Photo absorption peak

Cross sections for interaction

Silicon

CdZnTe
Summary of Crystalline Silicon Properties for Mammography

- Mobility electrons: 1350 cm²/Vs
- Mobility holes: 480 cm²/Vs
- Fano factor: 0.1
- Signal: 5500 electron-holes pairs for 20 keV x-ray
- Noise (dark current): ~1 electron
- Breakdown voltage: 3x10⁵ V/cm
- Fluorescence generation: No
- Dead time:
  - Deadtime per event: 189 ns
  - Deadtime at 20 kHz: 0.4%
  - Deadtime at 200 kHz: 3.7%

...is already a major industry standard and easily available in large production volumes.

Measured Effect of Dead Time

E. Fredenberg et al. SPIE 7258 (2009)
Charge sharing

How to avoid problems with charge sharing: "Co-incident Logic"

Effect of charge sharing

\[
\frac{NPS(u)}{\langle s \rangle} = \frac{(1 - \chi)K_u(x) + \chi K_d(x)}{G(S)(1 + \chi)} = \frac{G^1 + \chi[1 + 2\cos(2\pi u/p)]}{1 + \chi}.
\]

\[NPS(0) = \frac{G(1 + 3\chi)}{1 + \chi} \]

*D. Fredenberg et al., Submitted to Nuclear Instr Meth A*

DQE + clinical data
DQE for all different mammography detector materials (Se,Csl, Si)*

Note: DQE does not incorporate effects of energy weighting which benefits photon counting (10-15%)**

Elimination of electronic noise give low dose extra advantage for photon counting


Clinical Studies

dose, sensitivity and specificity

-75% dose reduction, increased cancer detection rate at same call-back rate

M. Wallis “Evaluation and clinical assessment of digital mammography screening using a Sectra Microdose full field digital x-ray unit and Sectra breast imaging PACS” NHSBSP REPORT 0601 MAY 2006 UNITED KINGDOM
-70% dose reduction, same cancer detection rate and call-back rate

Futu re applications: measuring the energy and using the spectral information
“Experimental evidence suggests that the growth of a tumor beyond a certain size requires angiogenesis”


Numerous Microvessels in Stained Section

Dromain et al. AJR 2006; 187:528–537

Typical MRI Images

Maximum-Intensity-Projection (MIP) contrast-enhanced T1-weighted coronal image

Multicentric invasive ductal carcinoma with extensive DCIS component (S:t Goran Hospital, Stockholm Sweden)

Contrast-enhanced Digital Mammography

Diekmann et al: 14-mm invasive ductal carcinoma (occult on mammography)
Challenges with Temporal Subtraction CEM

1. Compression may interfere with blood flow
2. Motion blurring because of long time between exposures (5-10 min)
3. Compromise in image quality of “standard” mammogram since the kVp is too high (40-50 kVp)

Dual Energy Contrast Enhanced Mammography (DE-CEM)

1. High Energy Exposure
   40-45 kVp, 200 mA, eg Rh anode, heavy filtering
2. Low Energy Exposure
   30 kVp, 140 mA, eg Mo anode
3. Combine the high and low energy images into one image
   Subtract the tissue, enhance iodine signal

Advantages: Less motion blur, compression not a problem

Spectral Imaging

Previous example: dual exposures
Photon-counting: single exposure

Non-overlapping spectra → better tissue cancellation
Concentrated around K-edge → improved iodine contrast
No motion blurring, easy to implement
Mammography is first with photon counting and thus a technology leader in medical x-ray imaging.

**Breast MRI v. Spectral Mammography**

**Spectral Mammography - a Future Alternative to Breast MR?**

- Same diagnostic information?
- Faster (no motion)
- Improved spatial resolution
- Similar appearance as normal mammograms
- Contrast to noise ratio better for MRI?
- 10 years more of clinical experience with breast MR

Easier in combination with biopsy?

- Cancer outside of the breast will not be detected

**Phantom study**

1-9 mm deep iodine containers in 15 mm PMMA
Anatomical clutter:
35 mm PMMA and oil
Previous study with a single detector line:

E. Fredenberg et al. SPIE (2008)
Dual Energy Subtraction

Logarithmic subtraction with 2 energy bins:

\[
\ln I = w \ln n_i + \ln n_i \quad \text{and} \quad \ln I_2 = w d_{i} d_{f} + K \\
S = \ln I_1 - \ln I_2 = w(d_{i} d_{f} + d_{i} d_{f}) \\
\Rightarrow S = 0 \quad \text{for proper choice of } w \quad (d_1 = d_2) 
\]

Cost of increased quantum noise?

Phantom study

Results with the full system:

- 40 kV, 3 mm Al, 1.13 mGy, ~10 s
- 6 mg/ml
- 9 containers visible in the subtracted image

Detector model

- Quantum efficiency, Intrinsic energy resolution of silicon
- Charge sharing: simulated**
- Pile-up: dead time measured to 247 ns
- Anti-coincidence logic: ~20% leakage
- ~150 ns AC window \Rightarrow chance coincidence
- Spread of thresholds: measured from threshold scans

*E Fredenberg et al., Submitted to Nuclear Instr Meth A
**M Lundqvist et al., IEEE Trans Nucl Sci, 2000
Detector model verification
Measurements of the PMMA-iodine phantom

Energy Resolution

- quantum noise dominates at high $f$
- anatomical noise at low $f$
**Generalized DQE**

\[
\text{GDQE} = \frac{\text{MTF}^2}{\text{NPS}_\text{quantum}}(\text{NPS}_\text{anatomical} + \text{NPS}_\text{quantum} - \text{MTF}^2)
\]


**Detectability index**

\[
d^2 = \int \int \text{GDQE} \cdot F^2 \cdot C^2 \, du \, dv
\]

~2.2 times improvement compared to the absorption image

**Can we use the new technology in mammography screening?**

Challenge

No contrast agent acceptable (?) except for high risk groups

But, spectral x-ray imaging may work anyway to improve the results in dense breasts

**Theory of Photon Counting**

\[
C_1 \frac{Z^4}{A} f_{\text{PE}}(E) + C_2 \frac{Zf_{\text{KN}}(E)}{A},
\]

- photoelectric effect
- Compton scattering

\( f_{\text{PE}} \) \( E^5 \), discontinuities at absorption edges

\( f_{\text{KN}} \) is the Klein-Nishina cross section

### Composition of the Female Breast (% of wet weight)

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Density (g/cm³)</th>
<th>Hydrogen</th>
<th>Carbon</th>
<th>Nitrogen</th>
<th>Oxygen</th>
<th>Ash (Si P K Ca)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fat</td>
<td>1.19</td>
<td>11.9</td>
<td>76.4</td>
<td>0.2</td>
<td>11.5</td>
<td>0.1</td>
</tr>
<tr>
<td>Skin</td>
<td>1.09</td>
<td>9.8</td>
<td>17.8</td>
<td>5.0</td>
<td>66.7</td>
<td>0.7</td>
</tr>
<tr>
<td>Adipose Tissue</td>
<td>0.93</td>
<td>11.2</td>
<td>61.9</td>
<td>1.7</td>
<td>25.1</td>
<td>0.1</td>
</tr>
<tr>
<td>Gland</td>
<td>1.04</td>
<td>10.2</td>
<td>18.4</td>
<td>3.2</td>
<td>67.7</td>
<td>0.5</td>
</tr>
</tbody>
</table>


### Elements in the Female Breast

<table>
<thead>
<tr>
<th>Element</th>
<th>Normal Tissue (median µg/g)</th>
<th>Malign Tissue (median µg/g)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Al</td>
<td>13</td>
<td>23</td>
</tr>
<tr>
<td>Br</td>
<td>22</td>
<td>35</td>
</tr>
<tr>
<td>Ca</td>
<td>294</td>
<td>653</td>
</tr>
<tr>
<td>Cl</td>
<td>10140</td>
<td>15225</td>
</tr>
<tr>
<td>Co</td>
<td>1.19</td>
<td>1.95</td>
</tr>
<tr>
<td>Cs</td>
<td>0.51</td>
<td>1.1</td>
</tr>
<tr>
<td>Fe</td>
<td>94</td>
<td>194</td>
</tr>
<tr>
<td>K</td>
<td>2620</td>
<td>14400</td>
</tr>
<tr>
<td>Na</td>
<td>9200</td>
<td>13230</td>
</tr>
<tr>
<td>Rb</td>
<td>11.5</td>
<td>28.5</td>
</tr>
<tr>
<td>Zn</td>
<td>3.5</td>
<td>57</td>
</tr>
</tbody>
</table>

15 paired samples


### Additional Elements in the Female Breast

<table>
<thead>
<tr>
<th>Element</th>
<th>Use</th>
</tr>
</thead>
<tbody>
<tr>
<td>Silicon</td>
<td>Breast implants</td>
</tr>
<tr>
<td>Iodine</td>
<td>Contrast agent (X-ray)</td>
</tr>
<tr>
<td>Gadolinium</td>
<td>Contrast agent (MR, X-ray)</td>
</tr>
</tbody>
</table>

### The HighreX Project

**Clinical Participants**
- Addenbrooke’s Hospital, Cambridge, UK (M. Wallis)
- Karolinska’s Hospital, Stockholm, Sweden (K. Leifland)
- Health Unit of Pietàs, Florence, Italy (M. Rossalli del Turko)
- Charitè Hospital, Berlin, Germany (F. Diekmann)
- Munster University Hospital, Munster, Germany (W. Heindel)
- Arcades, Marseille, France (B. Seradour)

**Aim**
Test 3D and Spectral Mammography (with and without contrast) versus standard mammography and breast MRI, enriched populations of 3*200 women

**Status**
2 units deployed with 3D imaging (around 200 cases collected)
1 unit deployed with spectral detector
Conclusions

• For mammography photon counting is out there in clinical reality (except for US)
• Low radiation dose advantage
• Spectral information most valuable for large objects (> a few mm)
• Promising results for contrast mammography with iodine
• Spectral information may be used to increase cancer detection
• Spectral information easy to implement as “add-on” information in mammography at almost no cost “to be used when desired”