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Advances in Computed Tomography and Digital Mammography

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ISMART, November 18, 2008, Kharkov

Outline

- **Advances in Medical X-Ray CT**
  - World market of X-Ray CT
  - Principles of third generation CT
  - Main performance parameters of modern CT
  - Data Measurement System in modern X-Ray CT
  - Future trends in X-Ray CT
- **Advances in Digital mammography**
  - Selenium/TFT technology
  - Performance of Selenium based mammography detectors
  - Manufacturing
  - Tomosynthesis

Global CT Market

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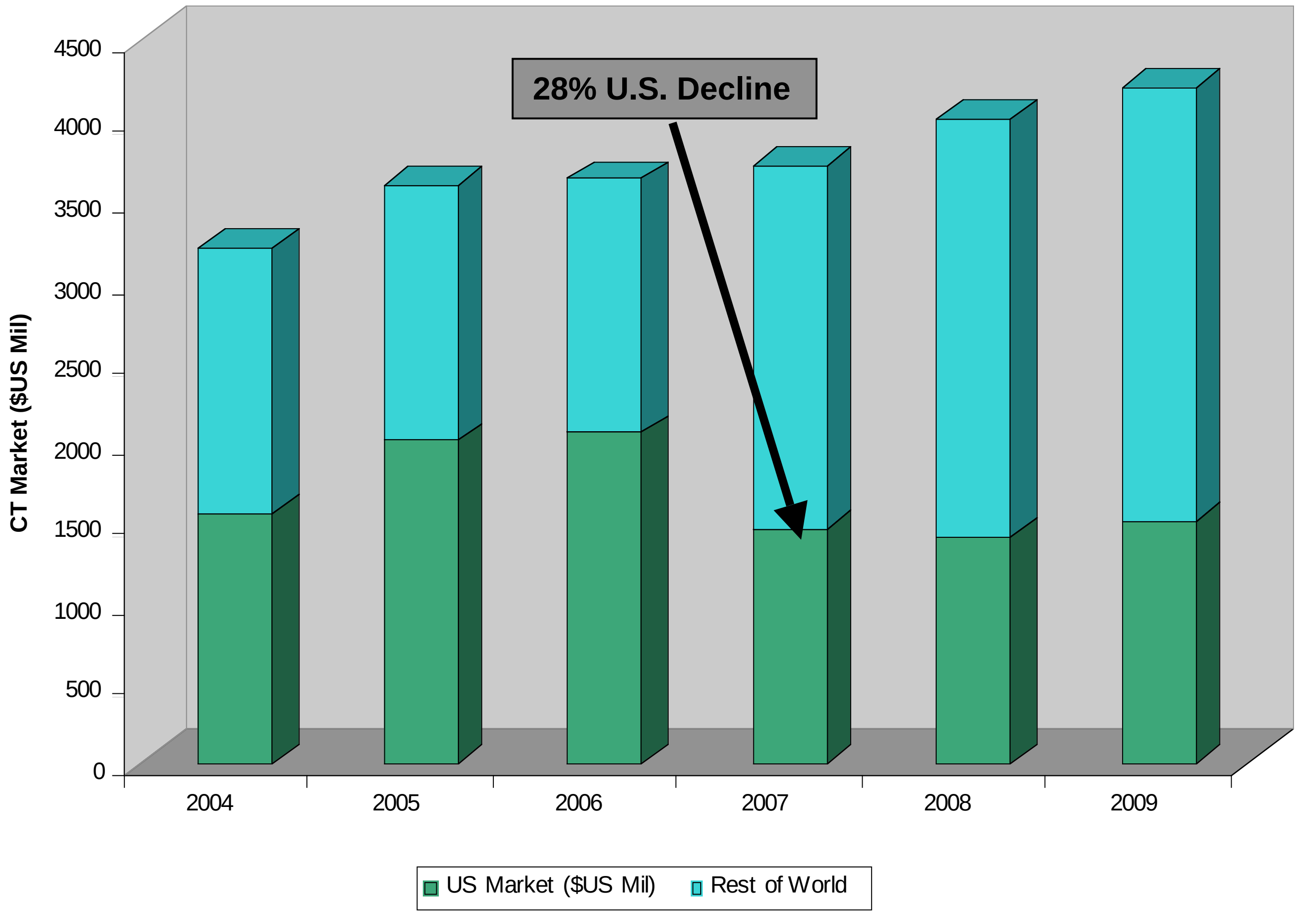
Note: U.S. data in orders



Source: Philips Medical, Fuji-Kezai, Analogic

US CT Market Distribution

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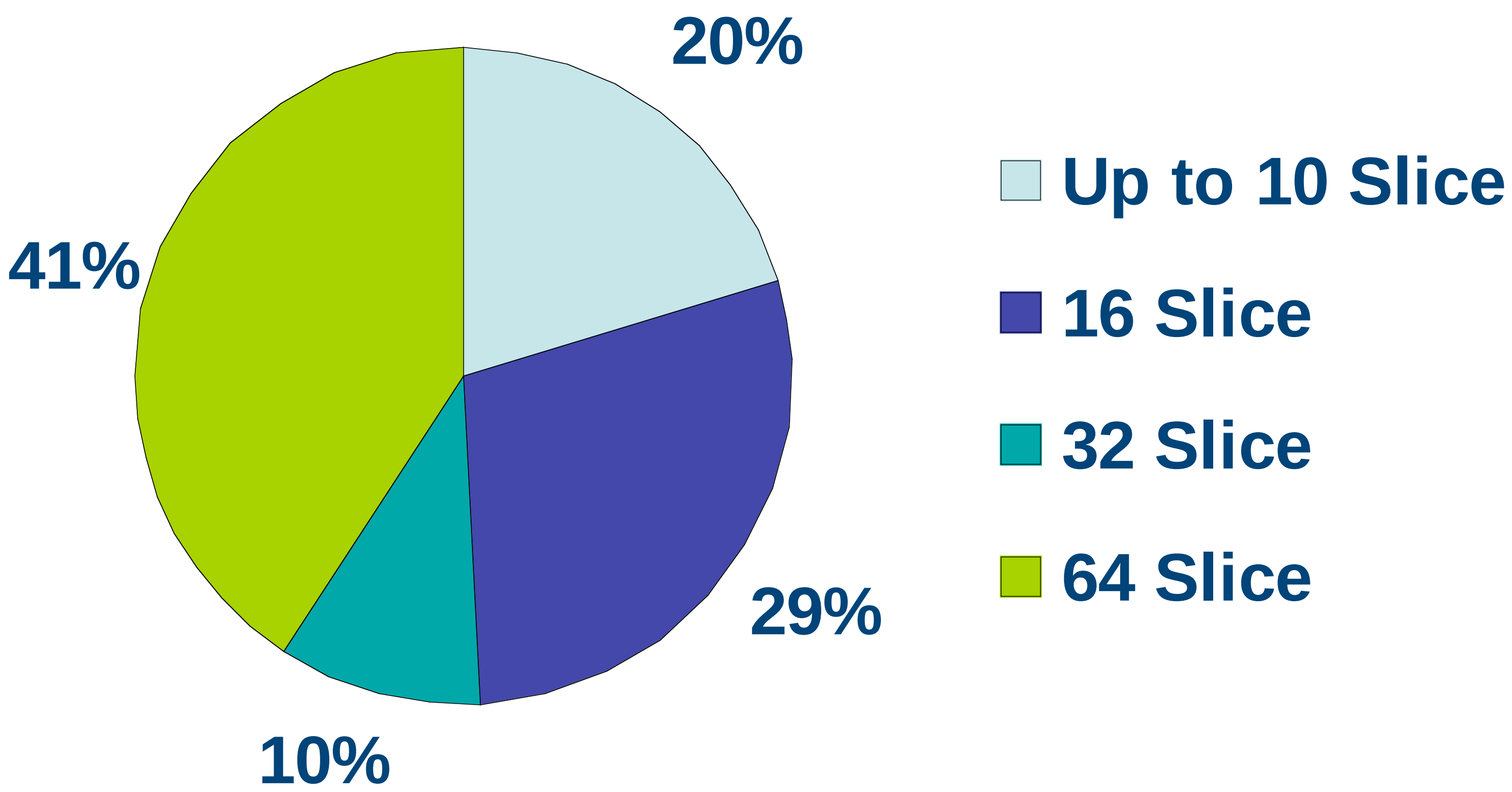
Market Share by Product Tier (2006) Based on unit volume



Source: 2006 Frost & Sullivan

CT Market Pricing Trends

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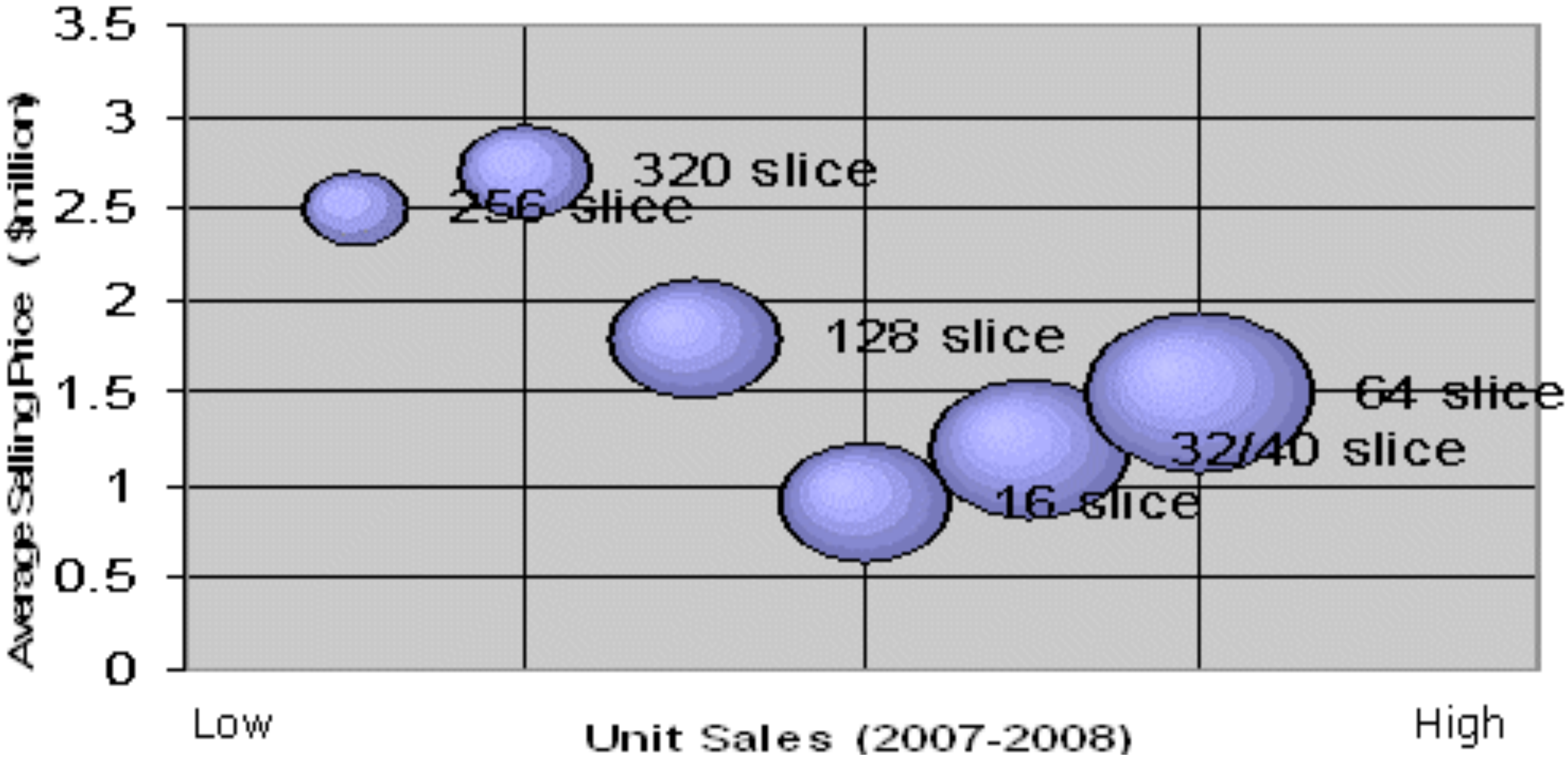

Source: 2008 Frost & Sullivan

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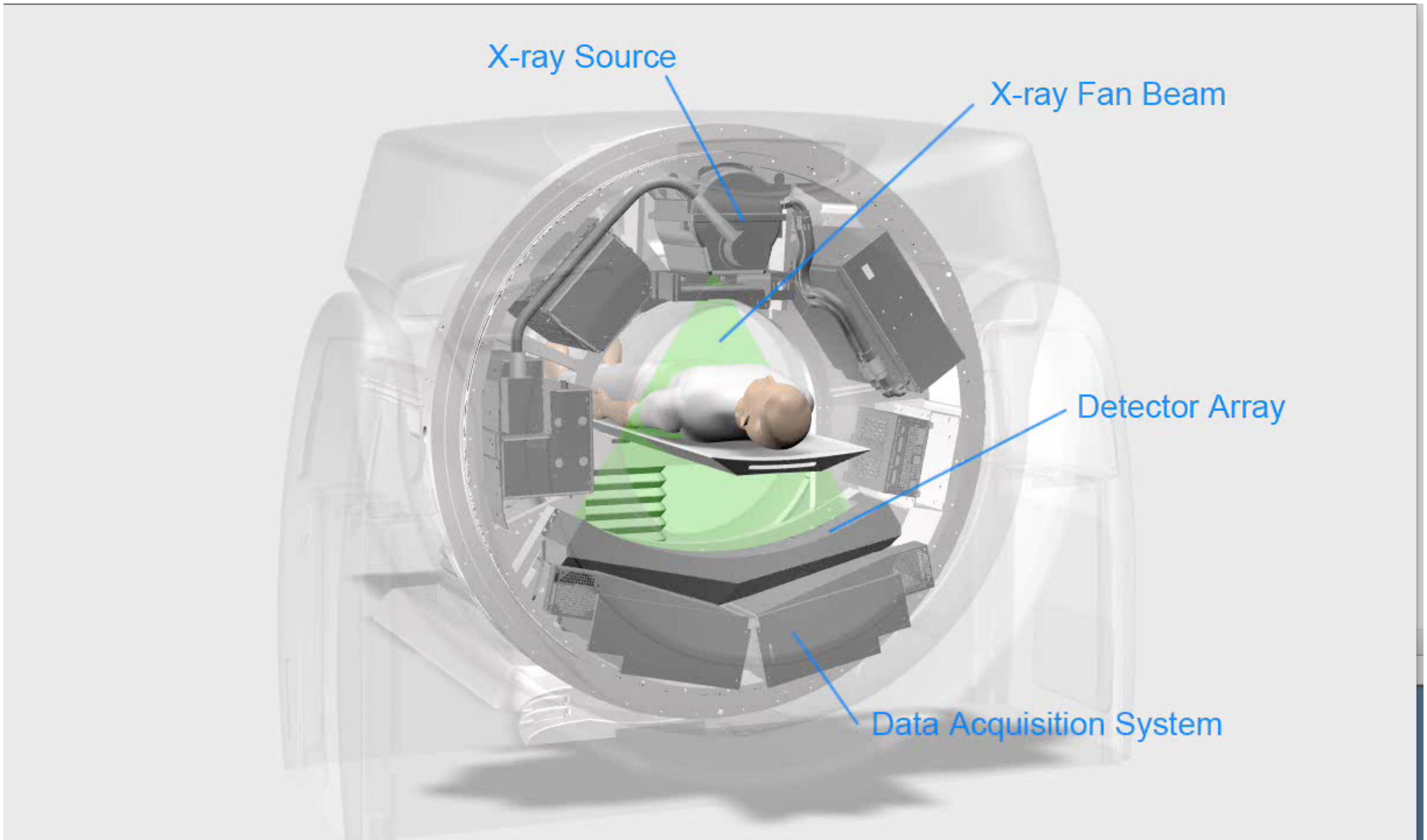
Evolution and Present Status of Medical X-Ray CT Imaging

Medical CT


X-ray Source
X-ray Fan Beam
Detector Array
Data Acquisition System

CT Systems Medical

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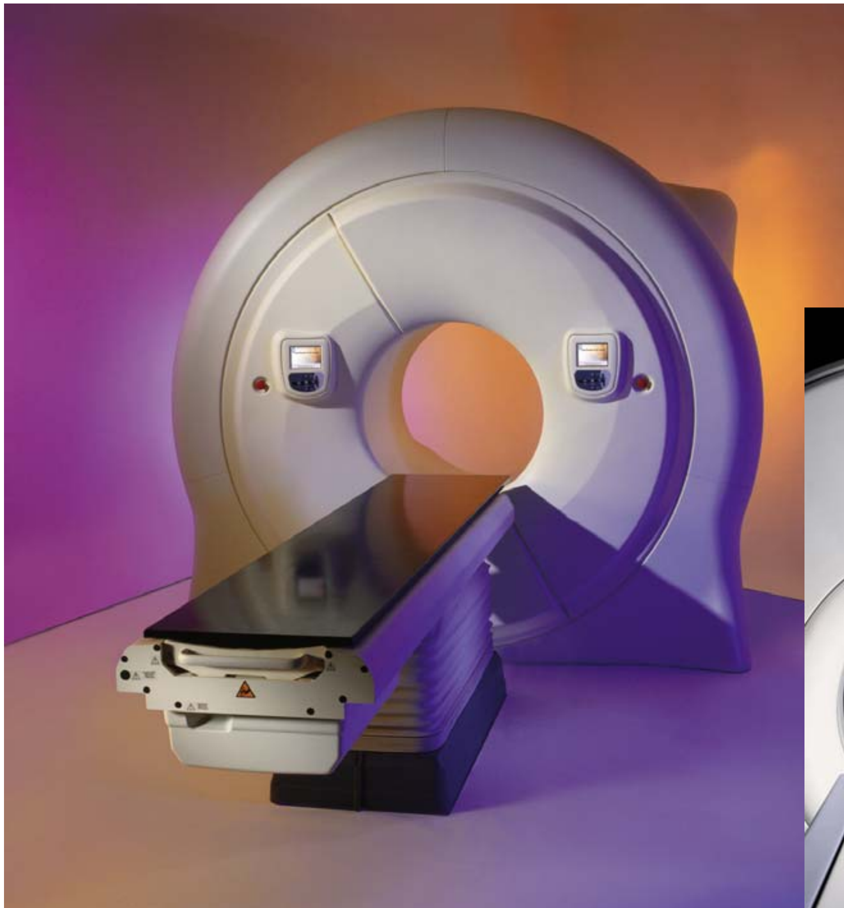

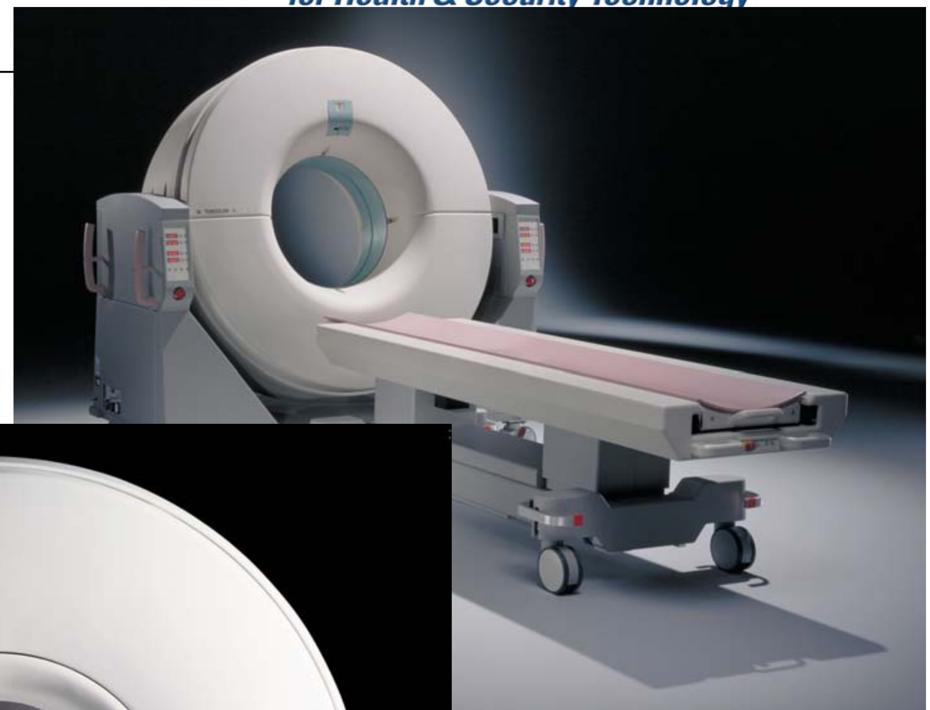

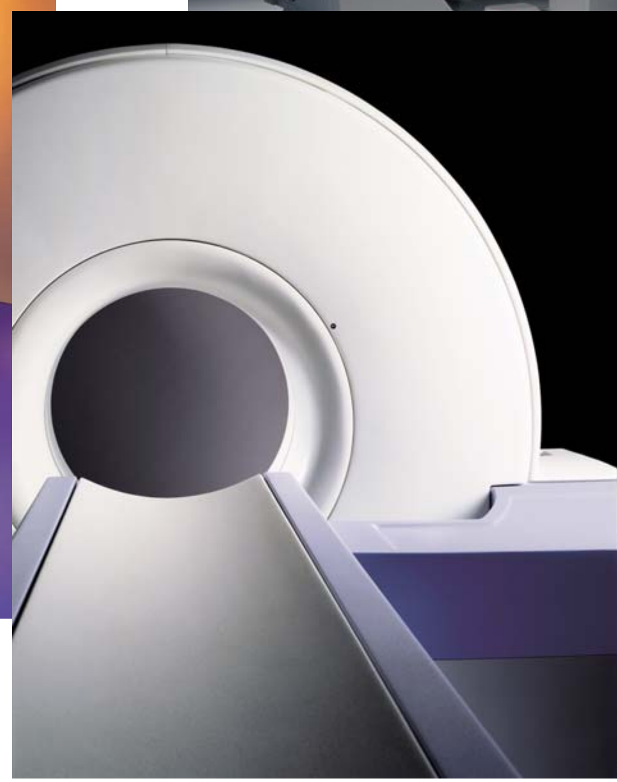

Analogic in CT

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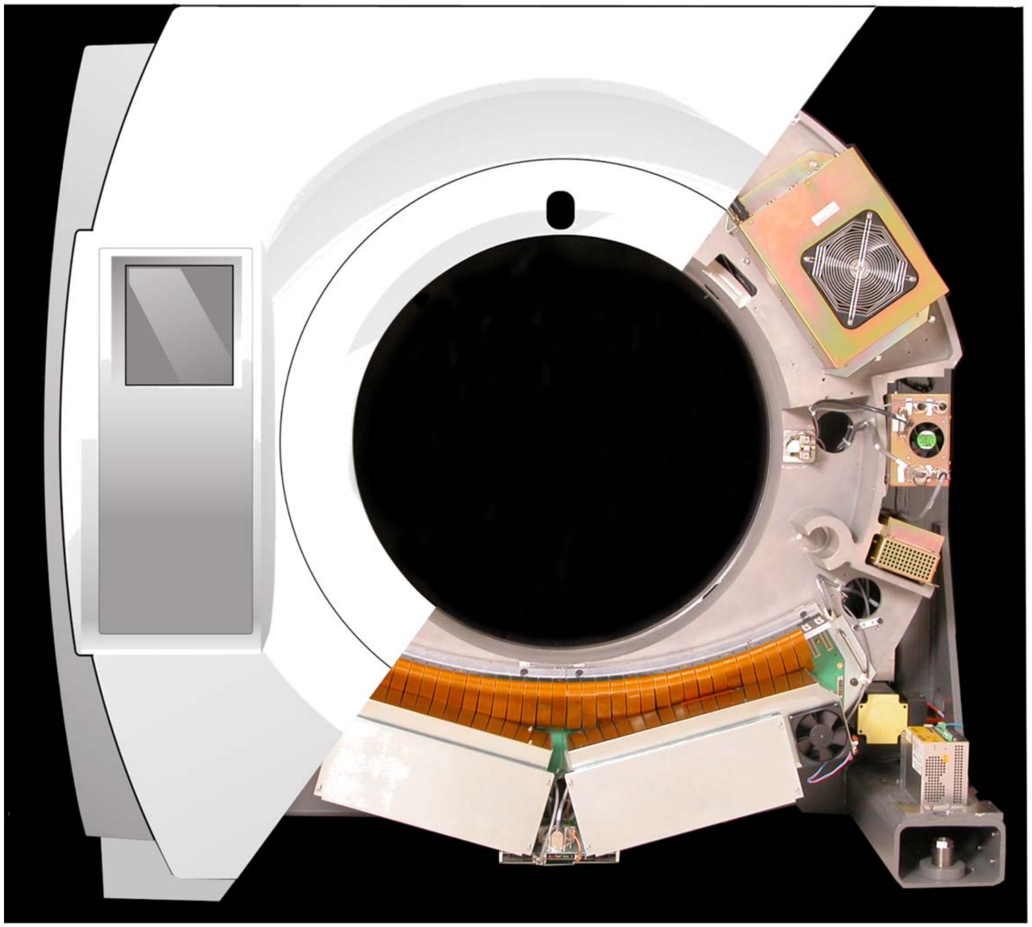
- CT Subsystems
  - Data Management Systems (DMS)
  - Data Acquisition Systems (DAS)
  - Detector Arrays
  - Gantries
  - PowerLink – Non-Contact Power Transfer
  - Collimators
  - Reconstruction Software
  - Motion Control
  - Patient Table
  - Operator Contol Station



CT Systems Security

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Milestones in X-ray CT

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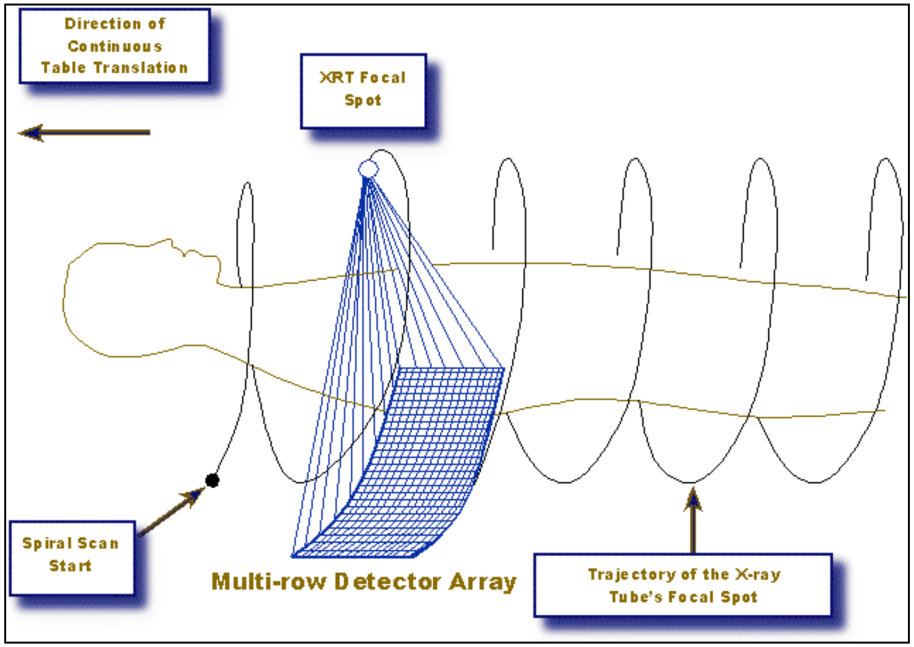
- **Sequential scanning of consecutive slices: 1970s**
- **Spiral scanning-CT acquisition with continuous translation of patient: beginning of 1990s**
- **Multislice spiral CT: end of 1990s**
- **Dual source CT: 2000s**

- ***Main trend: increasing speed of acquisition and axial coverage (slices)***

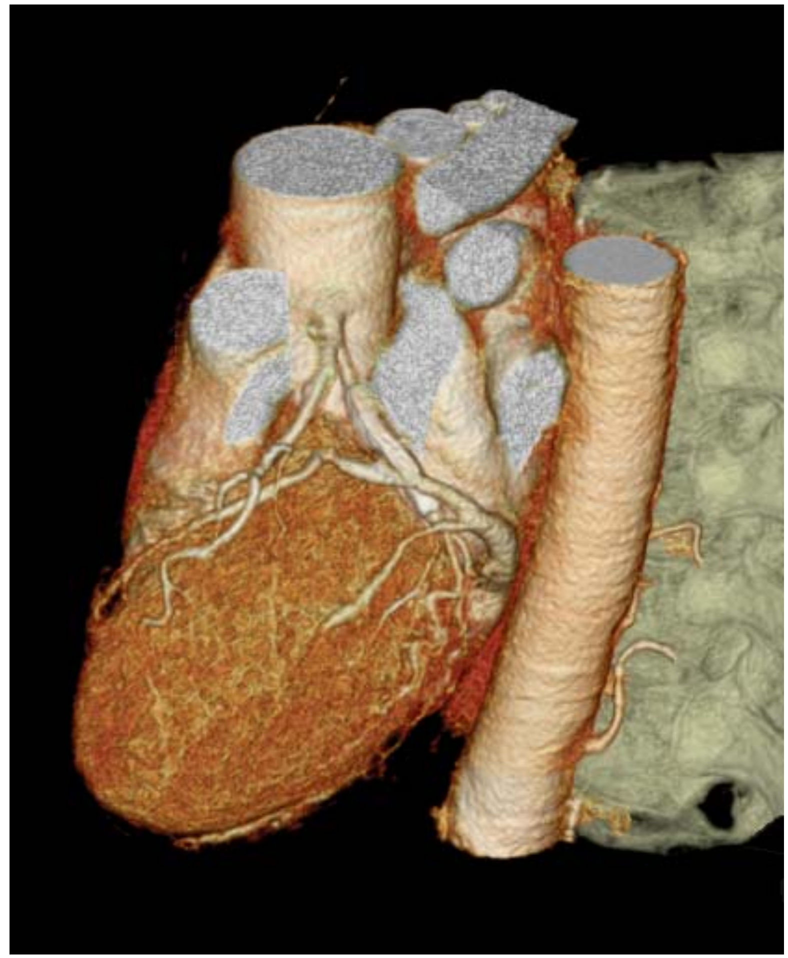
Multislice Spiral CT Scanning

- **Mid 1990s: 300 mm lung or abdominal examination with narrow slice requires 200 s long scan**
- **Mid 2000s: same examination takes 1-3 sec**
- **High speed of CT allows large scan within one breath hold, and to acquire images of moving organs, such as heart.**



CT applications

- **Coronary Angiography**
  - Rest phase of coronary arteries is 60 msec!
  - Non-invasive emergency diagnosis for cause of chest pain: coronary blockage, pulmonary embolism, aortic aneurism.
  - Alternative: 6 hour long invasive catheterization procedure with 1% risk of serious complications, including death.



Main drivers in X-ray CT

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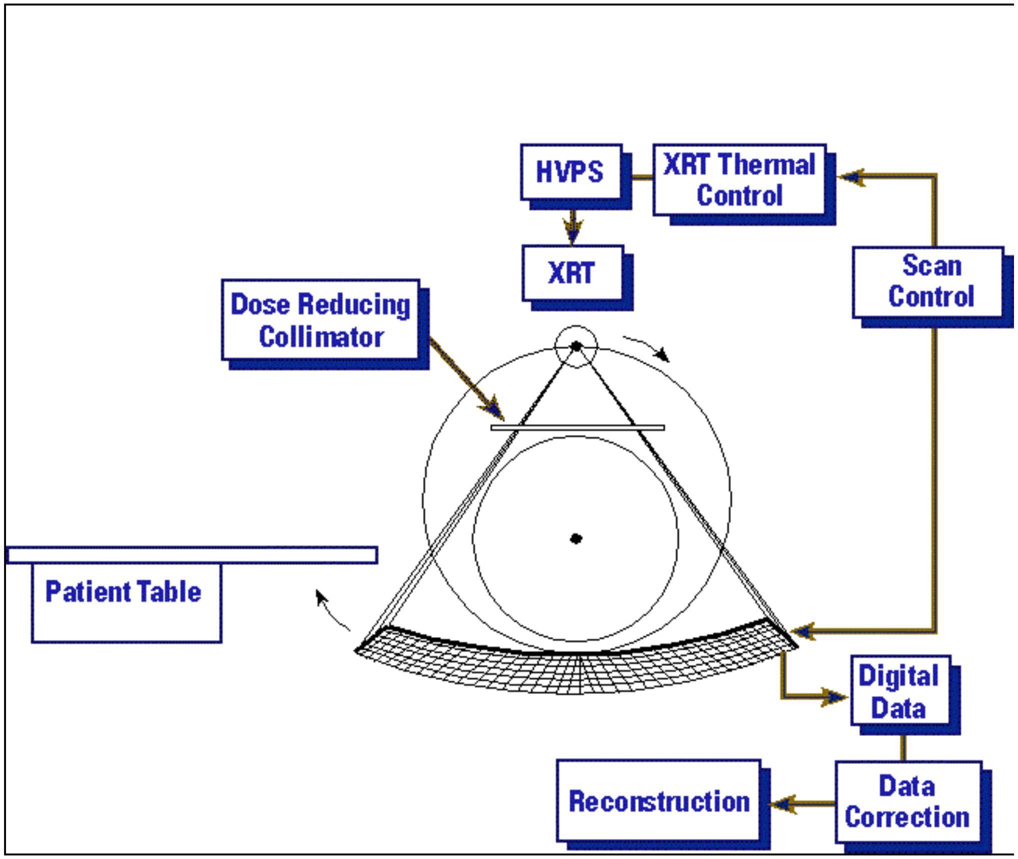
- Short scan time and large axial coverage for reliable anatomical and functional measurements of whole organs (perfusion of heart, brain, lung)
- Ultra-High spatial resolution
- Anatomical/Functional multi-modality imaging in SPECT-CT, PET-CT
- Patient dose reduction

- Is slice war over?

DMS for X-Ray CT

- **Scan time: 0.3 sec**
- **XRT Power: 100 kW**
- **Axial coverage: 40-160 mm at isocenter**
- **Spatial resolution: ~0.5 mm**
- **Remarkable progress in X-Ray CT in the past decade is largely explained by fast development of the Data Measurement System (DMS)**



Main DMS Parameters

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Parameter	Typical value	Parameter	Typical value
X-ray energy, kVp	80-140	Rotation time, sec	0.3
XRT power, kW	100	Data transfer rate, GBit/sec	10
FOV, cm	50	Operational temperature range	15-45°C
Isotropic resolution at isocenter, mm	0.5-0.6	Number of x-ray photons per sample at peak power	300,000
Number of channels per row	700-1000	Conversion efficiency, el/eV	0.01-0.02
Number of rows	64, 128, 320	Sampling rate, Hz	3000
Typical element size, mm	1x1	Data resolution, bit	18-20
Typical detector distance from XRT focal spot, mm	1000	Dynamic range, bit	16-18

Detector Channels per CT - 2008

High End (128-320 slices)	up to 300,000
Mid Range (16 slices)	57,00
Low End (single slice)	1000

Analogic DAS/DMS

- From 400-1000 channels in 1993 to 300,000 channels in 2008
- DAS/DMS complexity increases at constant:
    - Cost
    - Power consumption
    - and almost same Mechanical Envelope

Charge Integrating CT Detector

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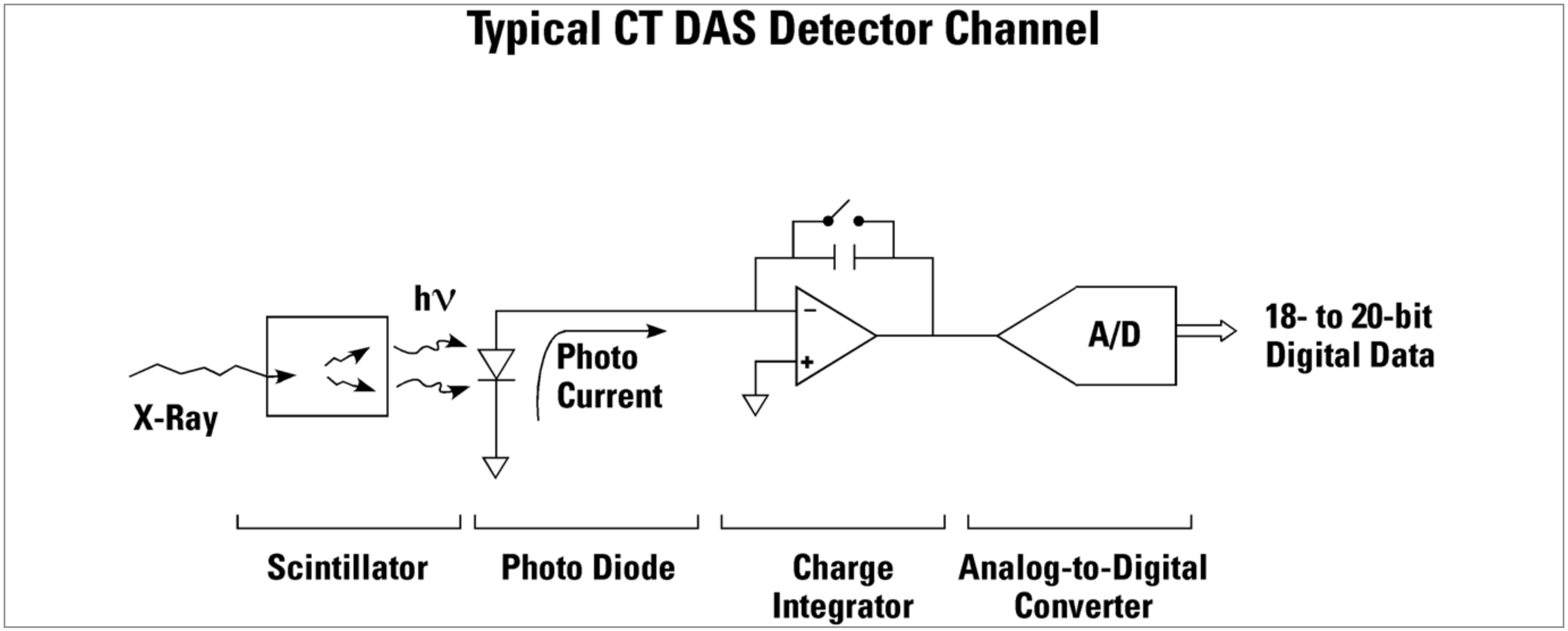

Uniform and Adaptive Detector Configurations

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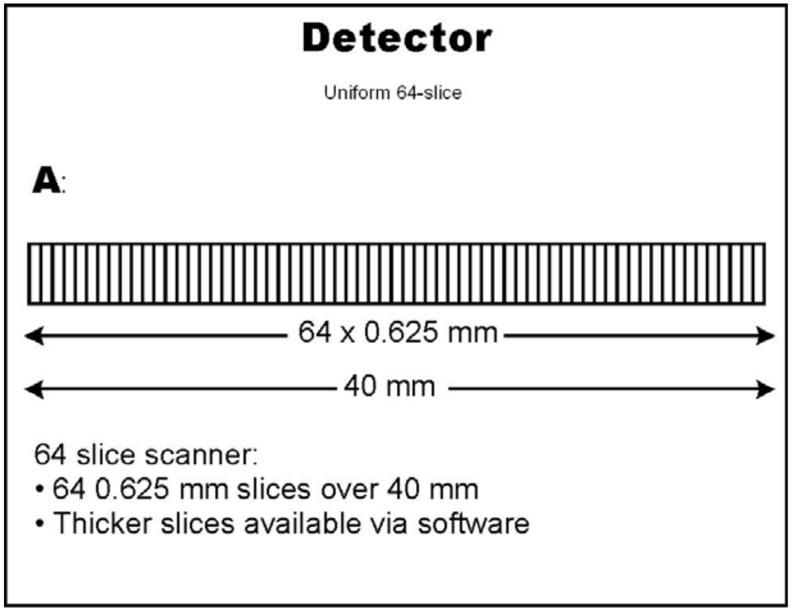

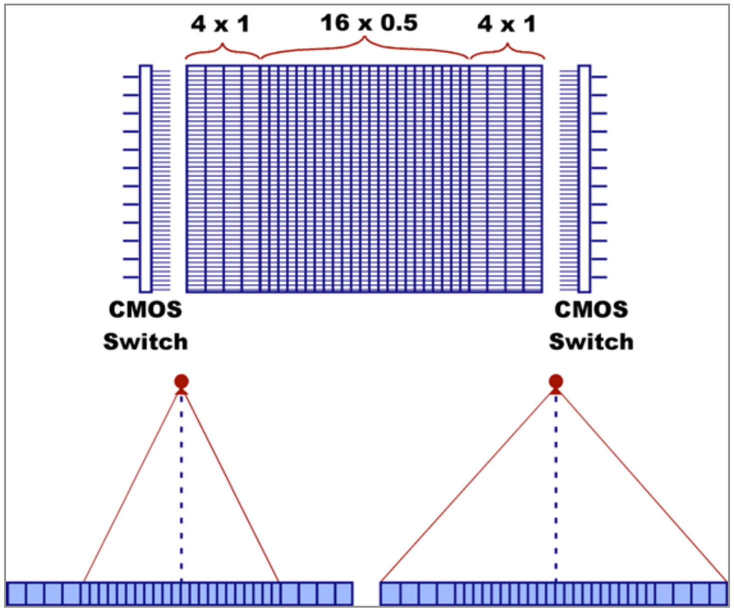

Uniform coverage in axial direction. Used in most 64, 128, 256 slice Scanners.

Adaptive arrays have fewer septa and DAS channels. Used in CT with 16 or fewer slices.

CT Subsystems

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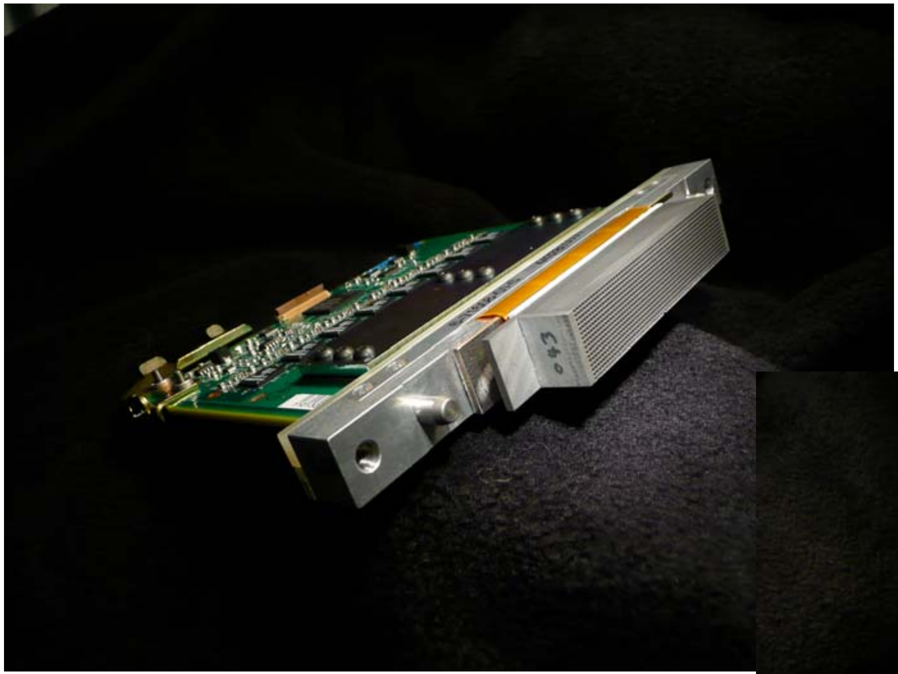
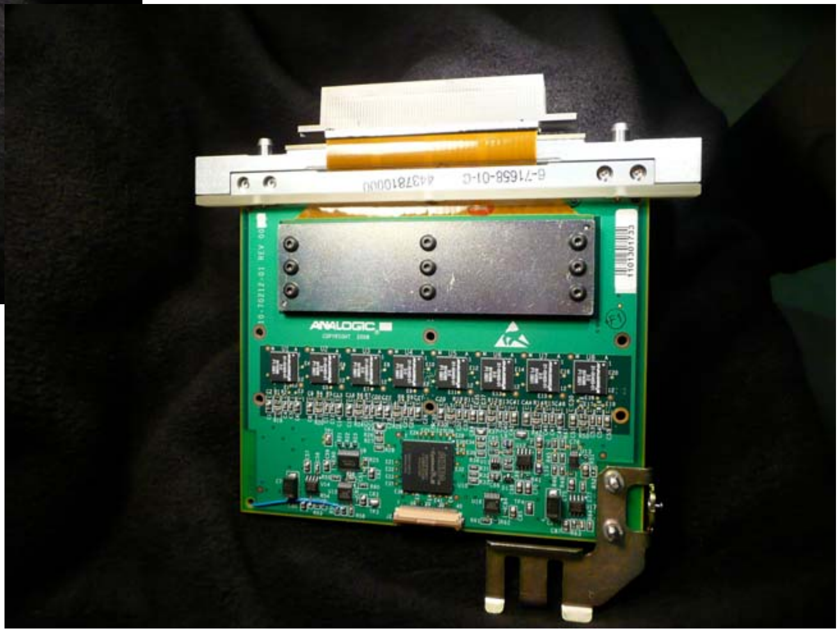
- **DMS**
  - 1 – 64 Slice+
  - Integrated DAS and Detector Assemblies
  - X-ray Beamline Design

16x64 CT Detector





Requirements for Scintillators in X-Ray CT

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Parameter	Requirement	Importance
Energy range	20-140 keV	
DQE (0)	>95 %	Image noise, dose reduction
Dose rates at detector	~0.1 Gy/s	
Lifetime dose	100 kGy	
Light output (LO)	>40,000 ph/MeV	Image noise at high attenuation
Emission spectrum, nm	400-1000	Used with Si photodiodes
Decay time	<10 μs	To support >10 kHz DAS rates
Afterglow	<10 ppm at 3 ms	Image artifacts
Susceptibility to radiation damage	<1 %/Gy	Image artifacts if channels non-uniform
Lifetime degradation	<20 %	Dynamic range reduction
LO temperature coefficient	<0.3 %/°C	Image artifacts if channels non-uniform
Low cost, non-toxic		DMS cost, cost of removal

Main Scintillators in X-Ray CT

Scintillator	Density (g/cm^3)	Thickness [a] to stop 99 % (mm)	Relative Light output[b]	Emission band maximum	Primary decay (μs)	Afterglow (% at 3 ms)
$CdWO_4$	7.9	2.6	1	495	2, 15	<0.1
$Gd_2O_3:Eu^{+3}$	7.55	2.6	-	610	-	-
$(Y,Gd)_2O_3:Eu$	5.9	6.1	1.52	610	1000	5
$Gd_2O_2S:Pr,Ce,$	7.34	2.9	1.8	520	2.4	<0.1
$Gd_2O_2S:Tb(Ce)$	7.34	2.9	1.8	550	600	≤0.7
$La_2HfO_7:Ti$	7.9	2.8	0.45	475	10	-
$Gd_3Ga_5O_{12}:Cr,$	7.09	4.5	1.38	730	150	<0.1

[a] Thickness to absorb 99 % of x-ray photons generated by tungsten anode x-ray tube at 140 kVp.

[b] Relative light output measured using silicon photodiode, under 140 kVp tungsten anode XRT excitation.

General Electric introduced fast "Gemstone" garnet based ceramic scintillator in 2007.
Limited data in public domain.

Silicon Photodiodes for X-Ray CT: Main Requirements

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Parameter	Value
Mode of operation	P-i-n structure, Photovoltaic, 0V bias
Elements number per chip, typ	512
Interconnect density	>64 per pitch
Pitch in x-axis, typ	1.0 mm
Pitch in z-axis, typ	1.0 mm
NEP	$<3\cdot10^{-15}$ W/$\sqrt{}$ Hz
Rise, fall time	<1 μs
Spectral response range	320-1060 nm
Photosensitivity, typ	0.3 A/W at 500 nm
Uniformity of photosensitivity	+/-2% ch-to-ch
Leakage current for 1x1 mm^2	5 pA, max,@10 mV bias, 25 C,
Cross-talk	0.1 %, max
Terminal capacitance for 1x1 mm^2	20 pF, max, @0V, 10 kHz

Data Acquisition Electronics: Main Requirements

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Parameter	Performance	Note
Electronic Noise σ_e	$\leq (1/2)\ \sigma_{\gamma,m}$	$\sigma_{\gamma,m}$ = Minimum Photon Noise
Digitization Interval δ	$\leq (1/6)\ (\sigma_{\gamma}^2 + \sigma_e^2)^{1/2}$	σ_{γ} = Photon Noise
Sampling Rate	≥ 10 kHz	
Offset Stability	≤ 1.5 ppm FSR /°C	FSR = Full Scale Reading
Gain Stability	≤ ± 50 ppm FSR /°C	
Integral Non Linearity	≤ 200 ppm R /°C ± 2 ppm FSR	R = Reading
Differential Non Linearity	≤ 30 ppm R /°C ± 1 ppm FSR	
Power Consumption	≤ 3 mW / Channel	
Packaging	64-128 Channels AMP/ADC	ASIC

Trends in Medical CT

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1. New CT scanning geometry: Dual-Source, Multi-Source, Inverse-geometry
- Advantages: Faster acquisition, Cone-beam artifact reduction
- Requires multiple DMS, or area detector, expensive

2. Energy-sensitive CT
- Advantages: elimination of beam hardening artifacts, material discrimination, better contrast at lower dose
- Solutions
  - Dual-layered detectors
  - Dual-source CT, kVp switching
  - Photon Counting Detectors with multiple energy bins

3. Multimodality CT: SPECT/CT, PET/CT, Preclinical systems

Trends in Medical CT

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4. Phase-contrast imaging (more distant future)

- Phase-contrast imaging, based on difraction is more sensitive in 10-150 keV range then attenuation based imaging.
- Requires interferometry and difractometry detection technique.

Multiple Source CT

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- **Multiple source/detector systems-old idea becomes a desirable development**
- **Higher rotation rates require increase in X-Ray power, not achievable with present X-Ray technology**
- **Fraction of rotation is required for a full scan**
- **Issues: scatter reduction, high cost**
- **Siemens introduced commercial Dual Source CT in 2004**

Energy sensitive CT: Dual Energy Detector

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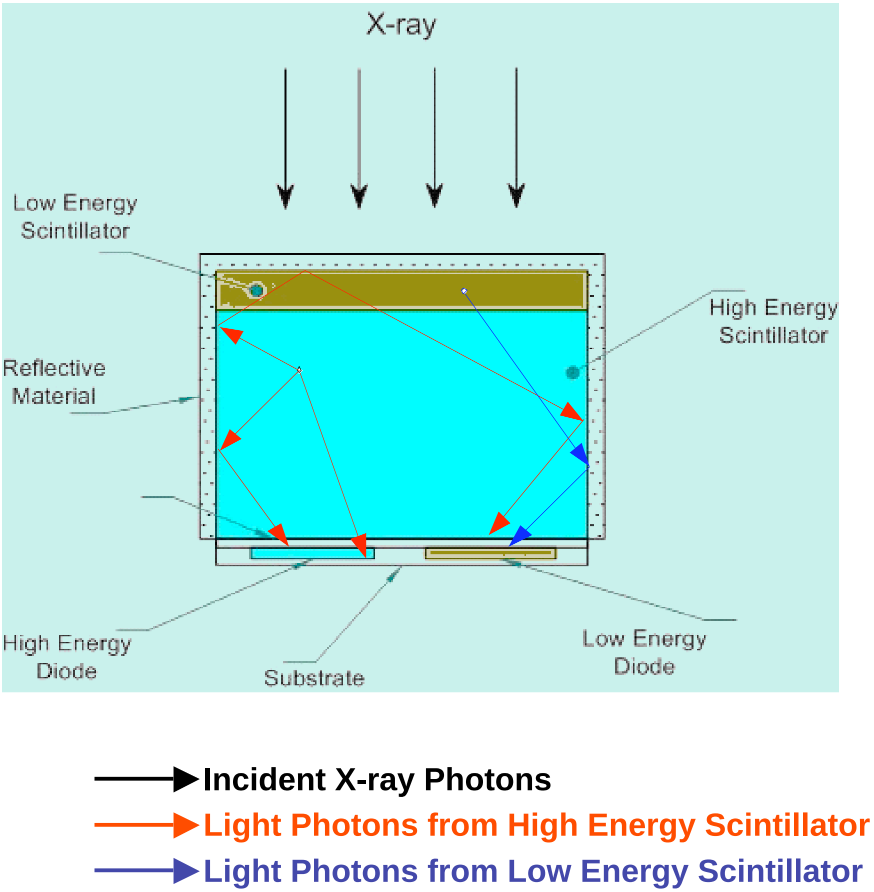
- **Two crystals with different emission bands are used.**
- **Radiation is hardened by the first crystal.**
- **Optical band pass filters limit diodes to see signal from only one crystal.**
- **Advantages:**
  - Simultaneous acquisition of Low and High Energy samples.
  - High Quantum Efficiency
  - Planar silicon PDA technology

R. Deych, US Patent 7,388,208 B2 2008



Contrast-to-Noise Model Results

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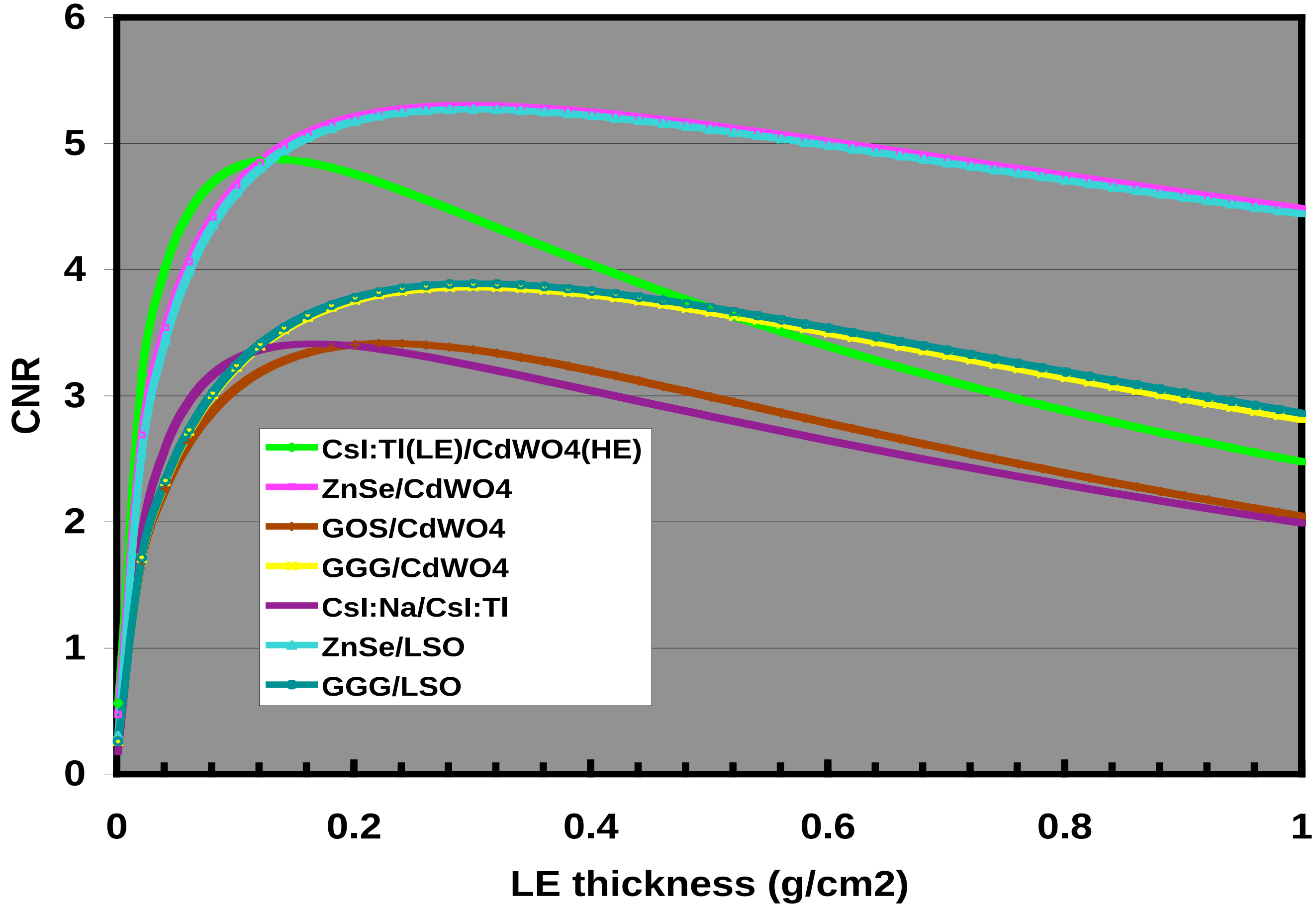
(Teflon detail in water background)



Energy sensitive CT, Single Photon Counting

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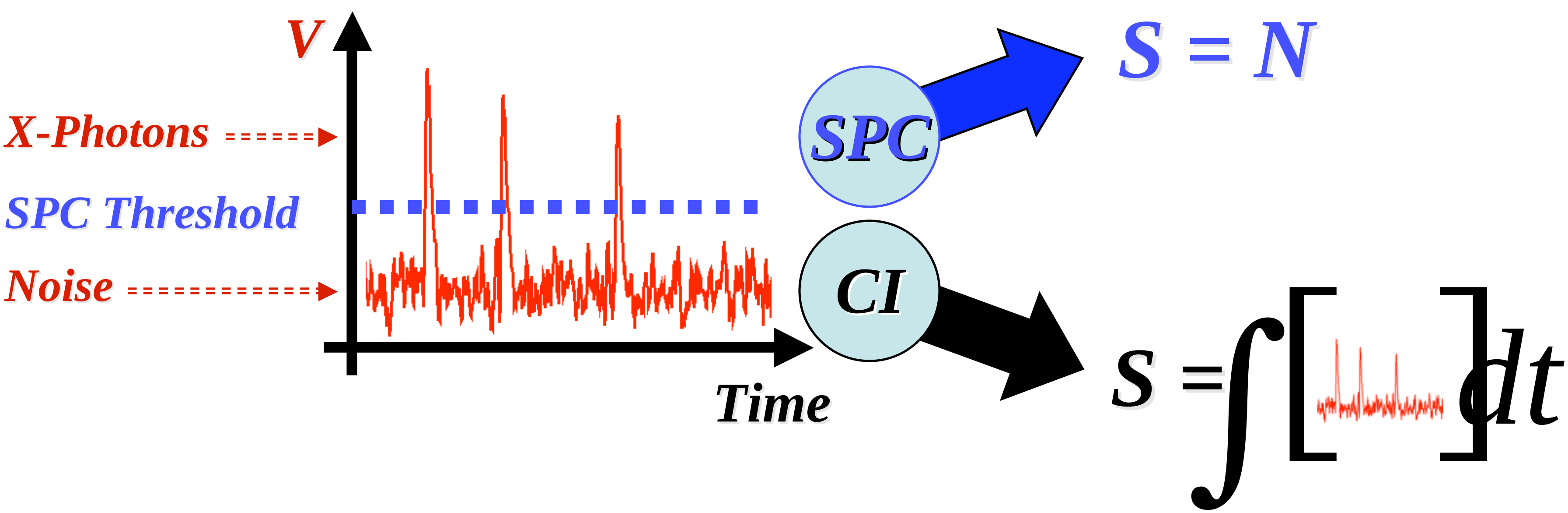
Single Pixel Analog Line



SPC in Computed Tomography

- **New medical applications and capabilities**
  - **Contrast media removal in images**
  - **Multiple contrast agents**
  - **Reduction of beam hardening artifacts**
  - **Patient dose reduction**
- **Requires high counting rates up to 10^9 (!) photons/sec/mm^2**

Direct Conversion Detectors

- **Dual energy CZT based detector tested in LightSpeed GE CT scanner, *IEEE 2007***
- **Pre-clinical CT scanner with 6 energy bands based on CZT technology tested by Philips, *IEEE 2008***
- **Main drawbacks:**
  - Long carrier transit time, insufficient speed
  - Material polarization at high exposure rates

Scintillator Based SPC

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- X-ray CT will require fast scintillators and internal gain in photodetectors
- Fast scintillators with Solid State PM are being proposed for CT
- Potential Available Fast Scintillators: LSO, LYSO, $LaBr_3$
- New faster scintillators with 1-10 nsec decay time are required

Multimodality CT: SPECT/CT, PET/CT

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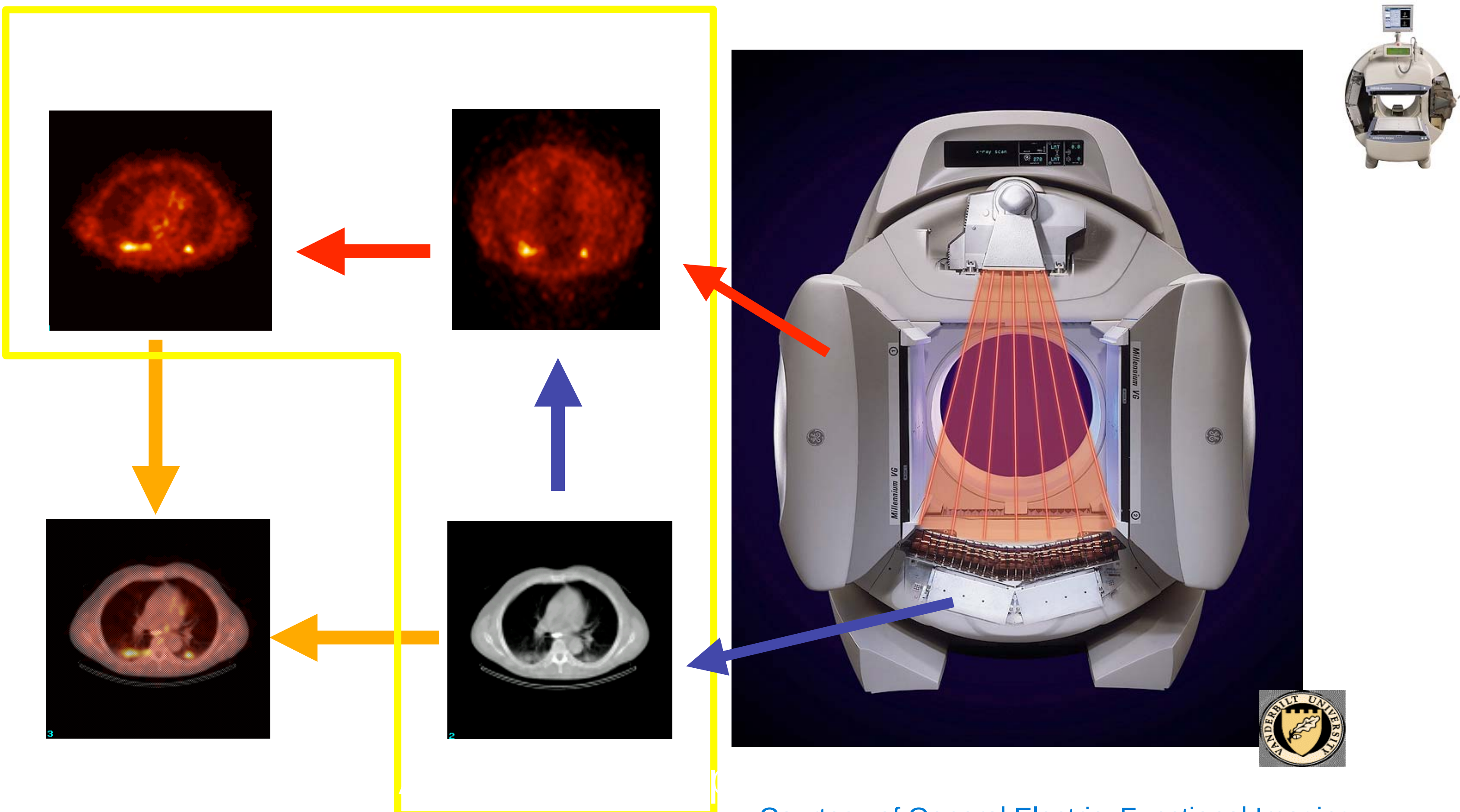

Courtesy of General Electric, Functional Imaging

Advances in X-ray CT: Conclusions and Predictions

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- **CT scanners with 320 slice acquisition in 0.3 sec are available**
- **The “slice war” between major medical imaging companies is over!**
- **New CT systems will include novel scanning techniques: multiple sources, inverse geometries**
- **Multi-energy CT will be needed to obtain better tissue discrimination at lower patient dose.**
- **Photon counting detection may replace charge integration**
- **X-ray CT will become new market for ultrafast nanosecond scintillators.**

Advances in Digital Mammography

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Stating the Problem

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Increasing mammography clinical diagnostic's sensitivity and specificity while optimizing patients' flow and reducing operational costs.

Digital Mammography Installed Units

2005	450
2006	900
2007	1650
2008	2000 est.
2009	2100 est.

Average Time/Patient

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Film-based analog mammography:

12 – 15 minutes

Se-based digital mammography:

5 – 6 minutes

Digital Radiography

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- **Two Step Conversion - INDIRECT**
  X-Ray to light to electrical charge

- **One Step Conversion - DIRECT**
  X-Ray to electrical charge

a: Se Technologies

- **Generation I:**

  dielectric isolation layer deposited on top of two layer "p" Se structure

- **Generation II:**

  single Se deposition process with real time doping to create three layer pin or nip structures

Mammography Detector

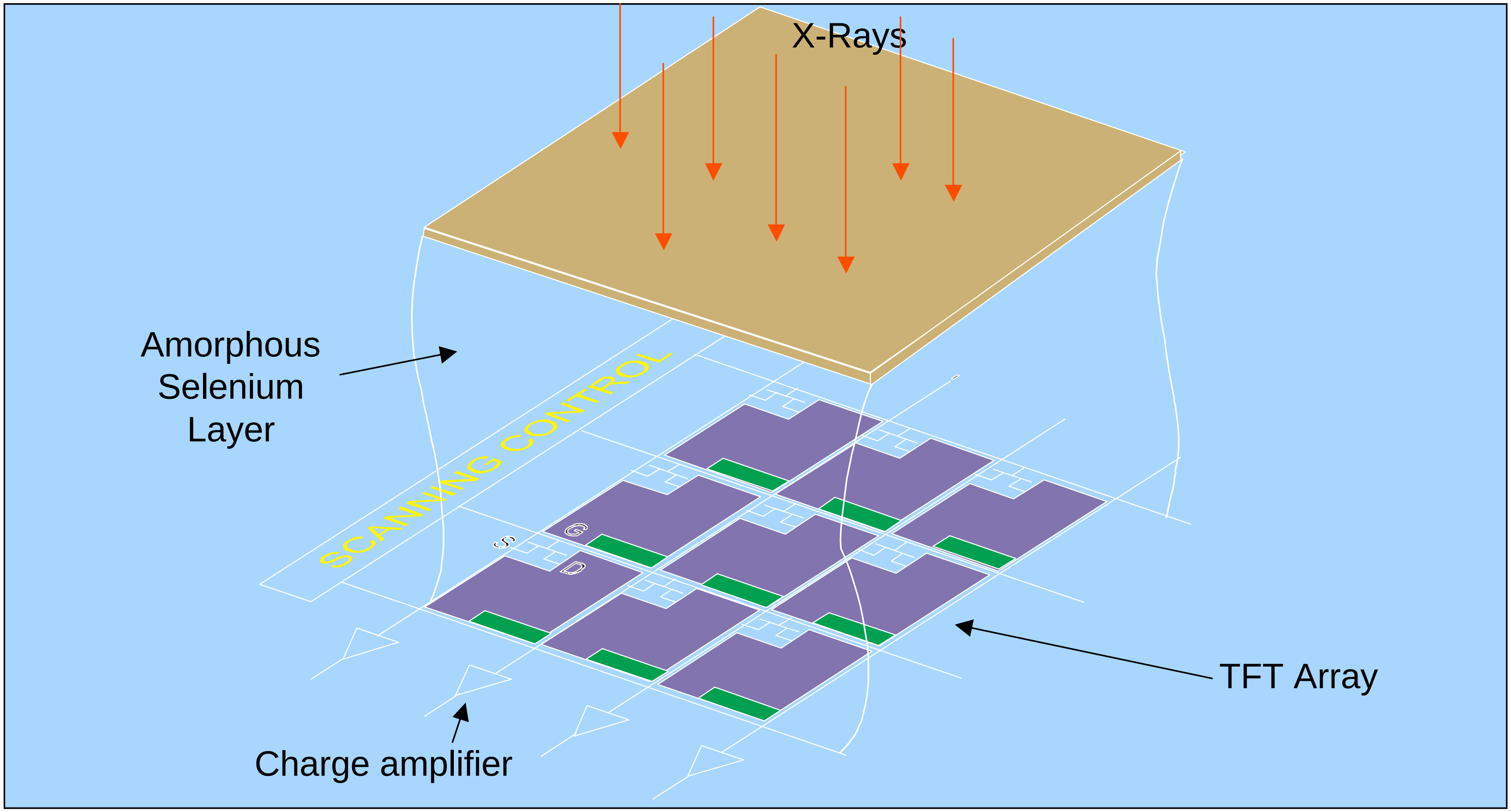
General Characteristics

Technology:	amorphous Selenium
Active area:	24 cm x 30 cm
Resolution:	2816 x 3584 pixels
Pixel pitch:	85 µm
Acquisition speed:	2 frames/second
Digitization:	14 bits

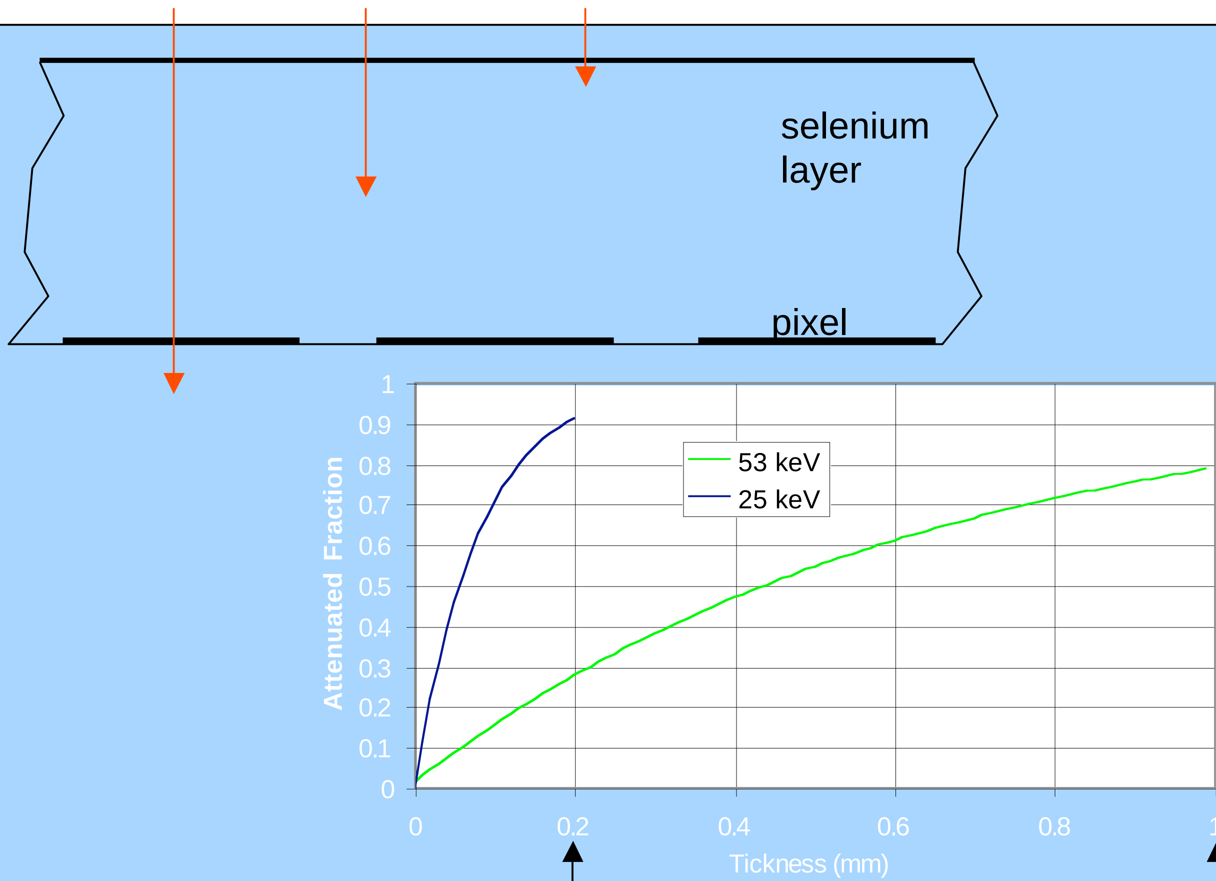
Se Characteristics

- **Atomic Number: 34**
- **Conversion Efficiency: 50 eV / e-h**
- **Evaporation Temperature: 217 deg. C**
- **Crystallization Temperature: 60 deg. C**
- **Expansion Coefficient: 40 ppm/deg. C**

a: Selenium Detector Structure



X-Ray Absorption



Mammography
25 keV < Ex < 40 keV

RT, Radiography
40 keV < Ex < 120 keV

Attenuated X-Ray fraction or Quantum Efficiency:

1 – exp[-aL]

α= a (E,Z,r)
mostly photoelectric

Charge Generation

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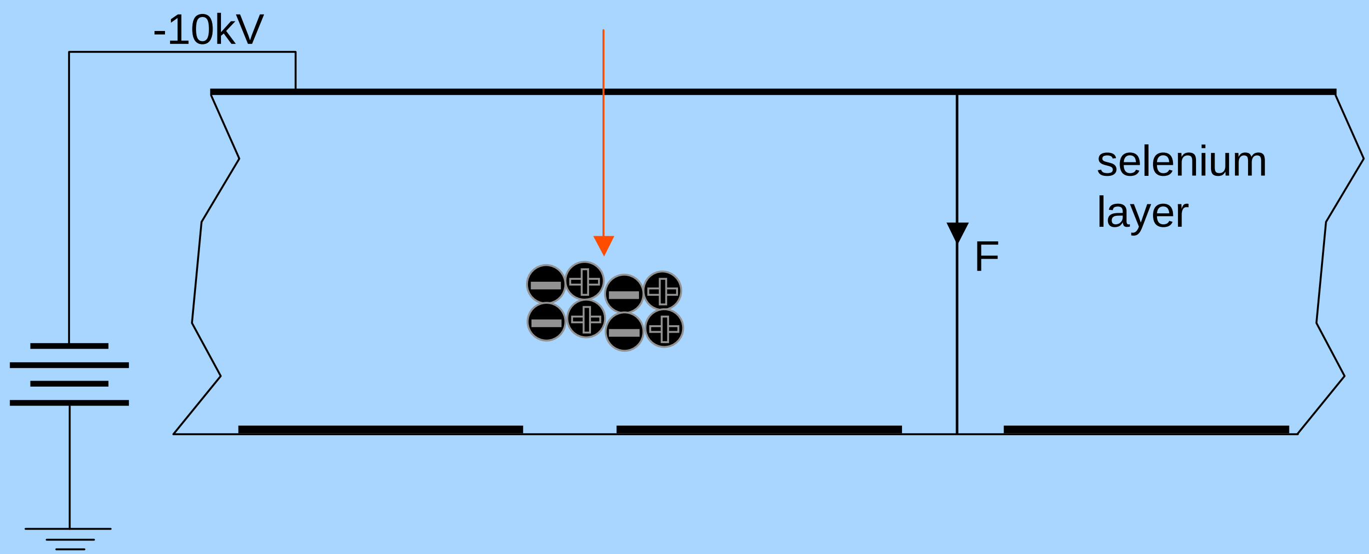

Number of electron-hole pair created:

$Ex / W_{+/-}$

$W_{+/-} = W_{+/-} (E,F)$

for $F = 10$ V/mm, $W_{+/-} \sim 50$ eV

Charge Drift

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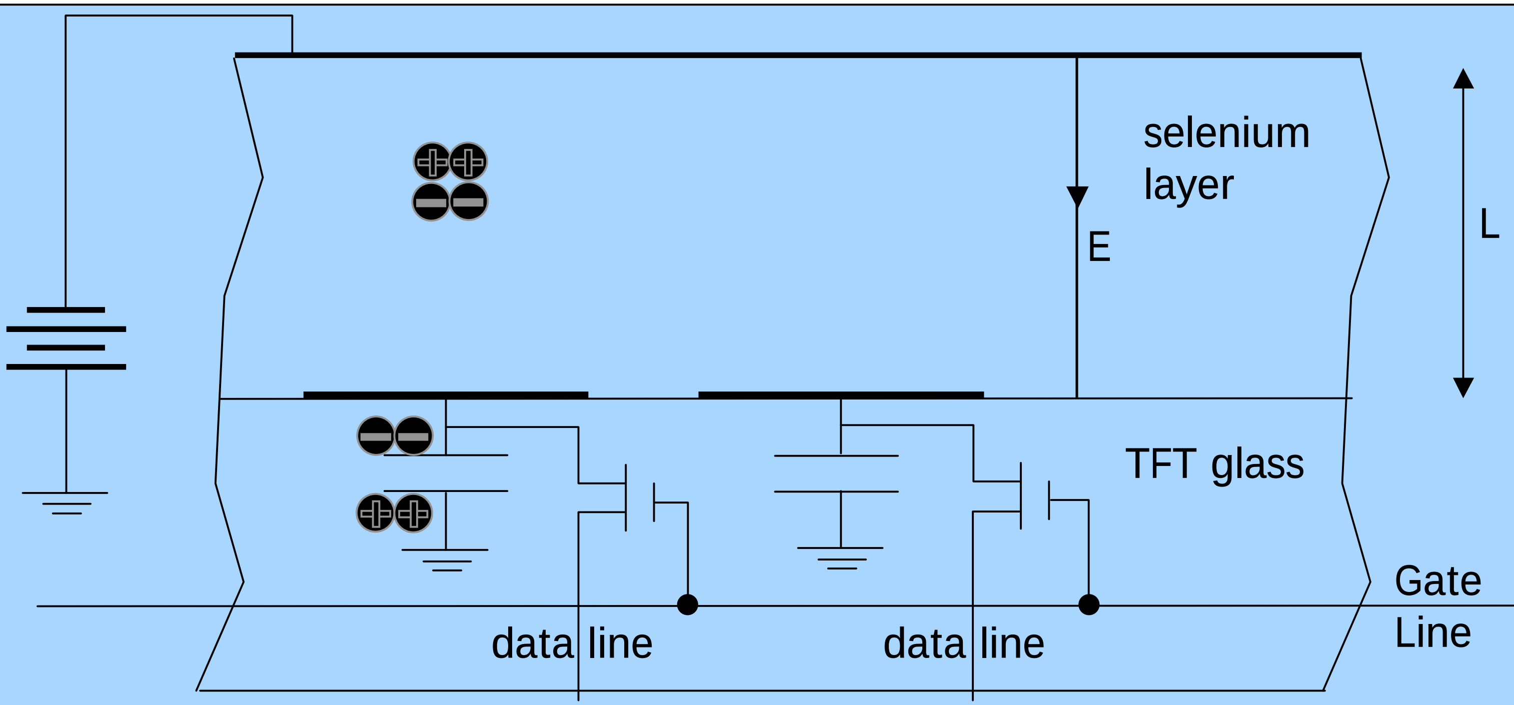


Charge Collection (induction) efficiency:

$\eta = \mu\tau E/L \{ 1-\exp[-L/\mu\tau E] \}$

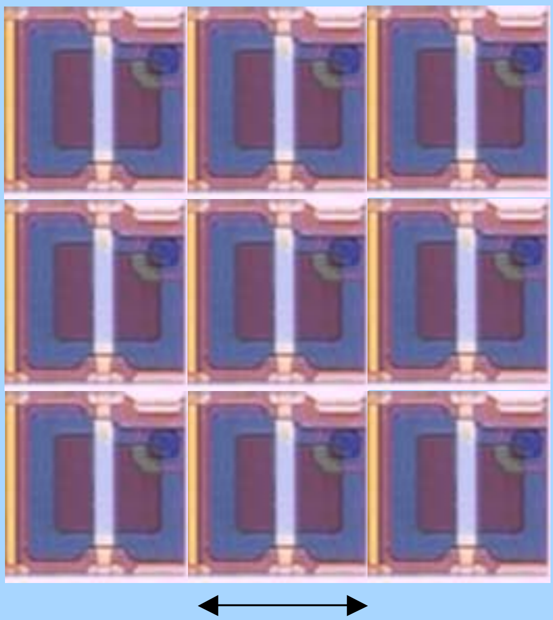
μ: mobility
τ: lifetime
$\mu\tau E$: mean free path

Typical values for a-Se:

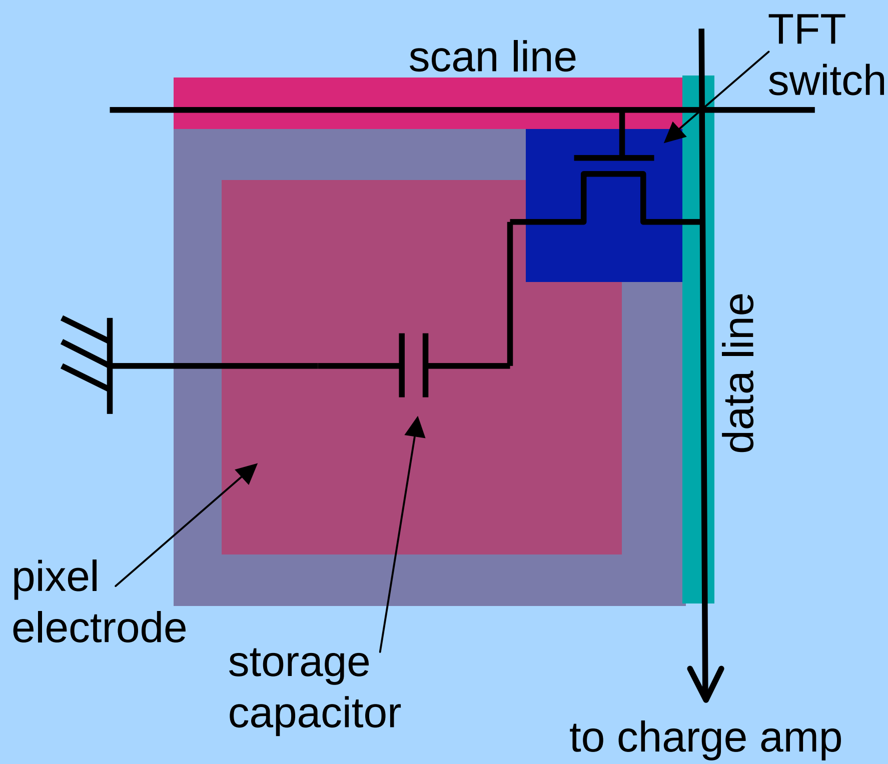
$\mu\tau E(e)$ = 3-4 mm
$\mu\tau E(h)$ = 3-20 mm

For a good detector $\mu\tau E >> L$

TFT Pixel Architecture

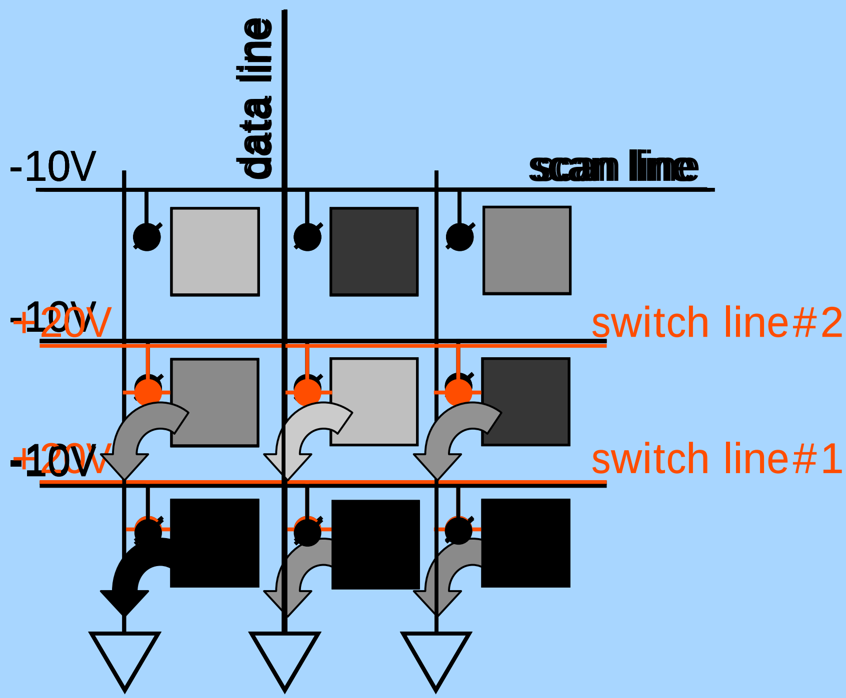


150 um: Real-Time, GR
85 um: Mammography



- Pixel pitch is larger than the pixel electrode (geometrical fill factor)

TFT Array Sequential Readout



Detector Spatial Resolution

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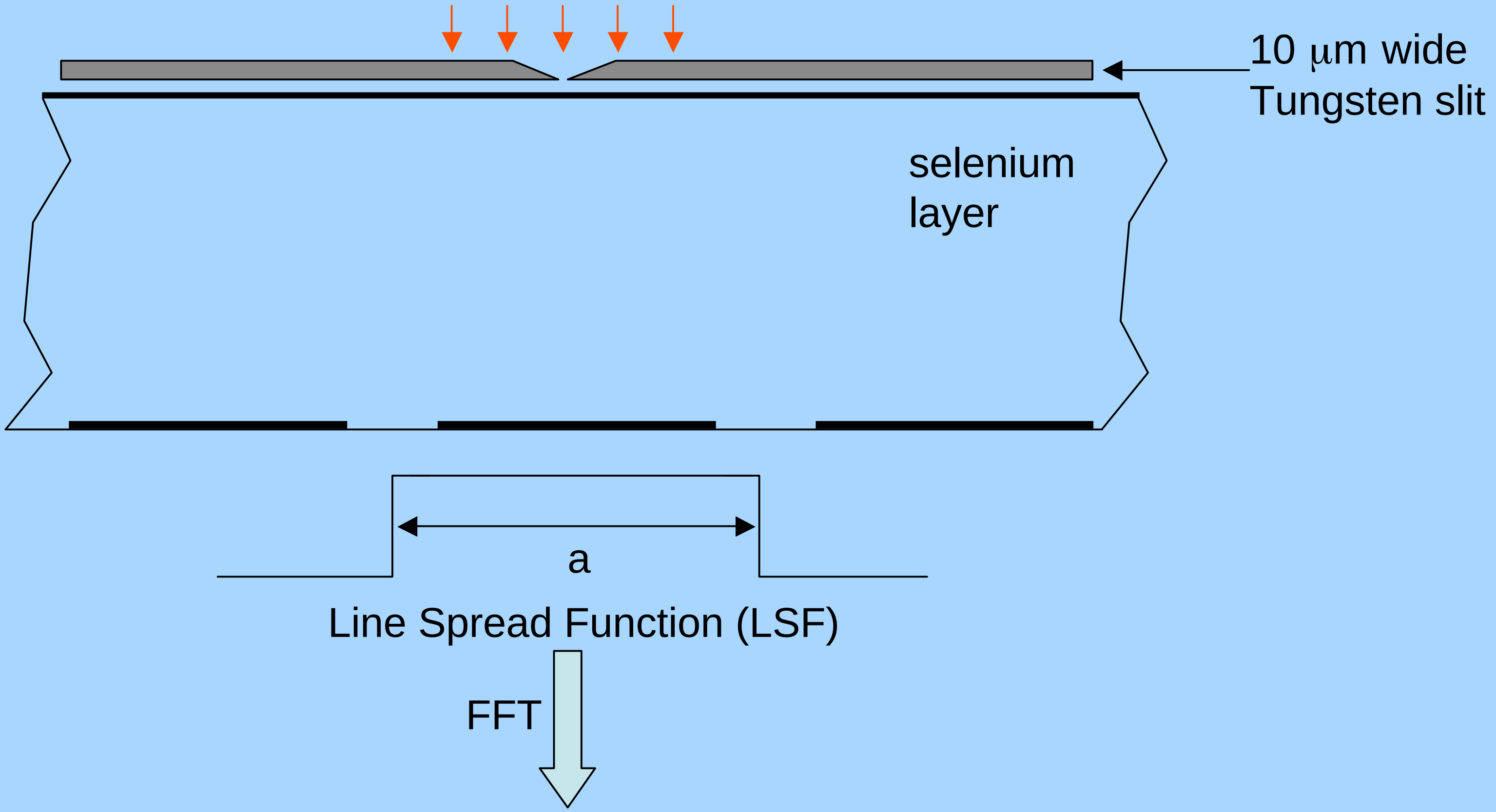

$$\mathrm{Sinc}(a,f) = \frac{\mathrm{Sin}(a\pi f)}{(a\pi f)}$$

Detector Modulation Transfer Function (MTF)

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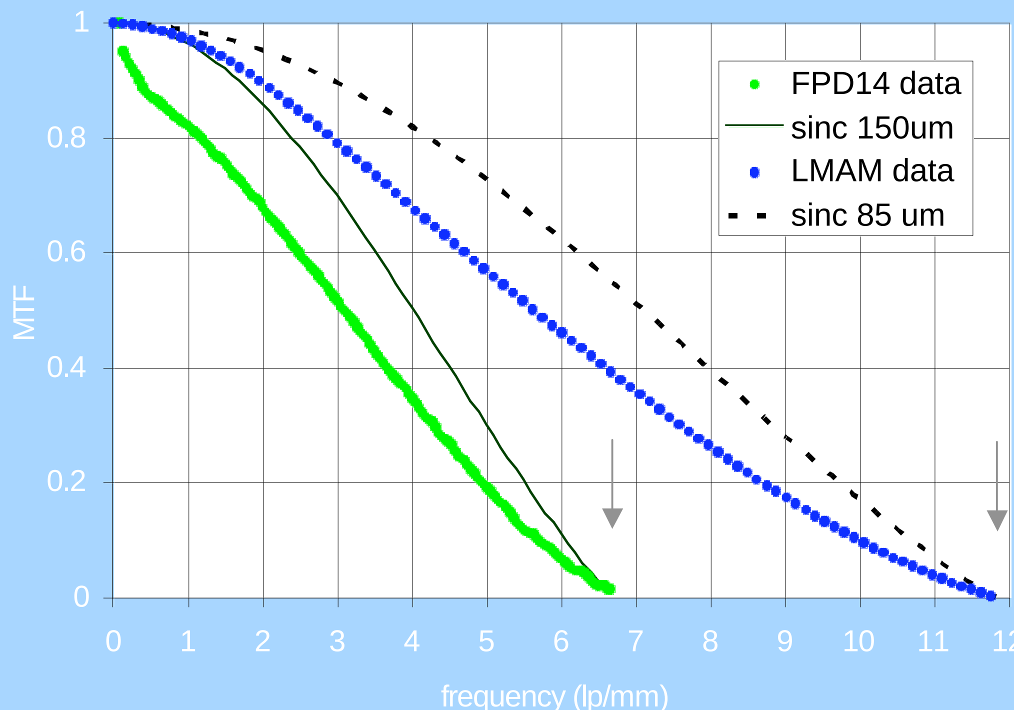

- Experimental MTF’s are close to their corresponding sinc functions
- The first zero-crossing of each sinc function corresponds to physical pixel pitch:
  6.6 lp/mm for 150 μm pixel
  11.7 lp/mm for 85 μm pixel

Detector Performance: Signal to Noise Ratio (SNR)

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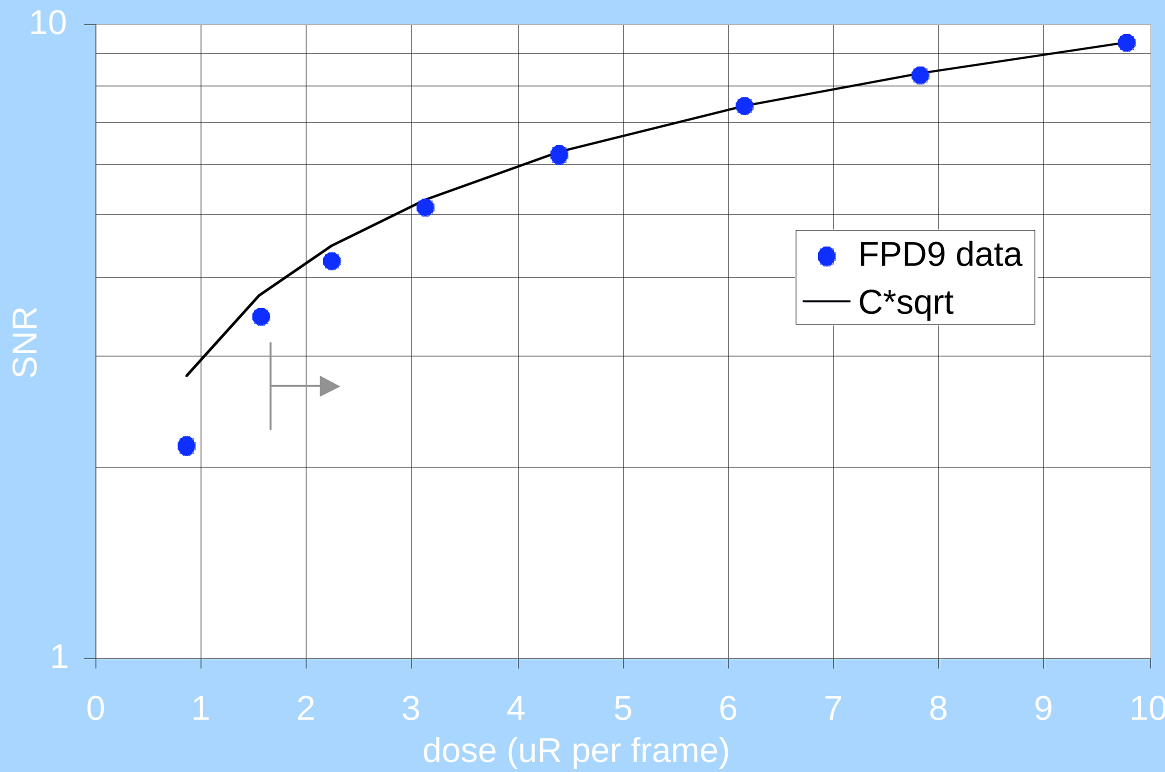
If γ is the number of X-Ray incident on the detector then where N_e is the electronic noise

$$SNR = \frac{\gamma}{\sqrt{N_e^2 + \gamma}}$$



The SNR curve follow a sqrt behavior above 2μR

Detector Detective Quantum Efficiency (DQE)

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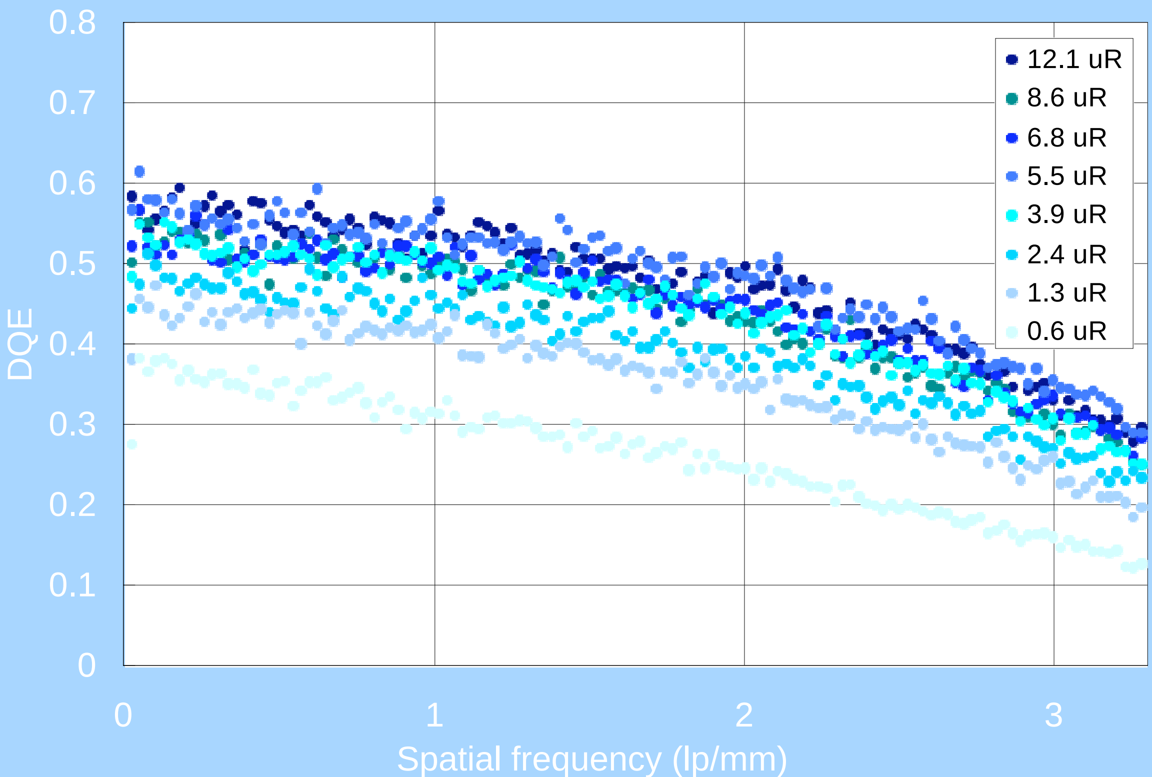

DQE remains high for high frequency values

$$DQE = \frac{(SNR)^2_{det}}{(SNR)^2_{in}} \xrightarrow[\text{domain}]{\text{frequency}} DQE(f) = \frac{G^2 MTF^2(f)}{\Phi \times NPS(f)}$$

Φ: X-Ray fluence
NPS: noise power spectrum
G: conversion gain

Manufacturing Steps

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- Deposit in vacuum amorphous Se on TFT
- Deposit top metal electrode on a-Se the
- Attach high voltage contact to electrode
- Deposit isolation on multi-layer structure
- Attach peripheral ASIC electronics to TFT
- Assembly the detector in final enclosure
- Perform parametric and imaging tests

Selenium Coater

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Selenium Coater

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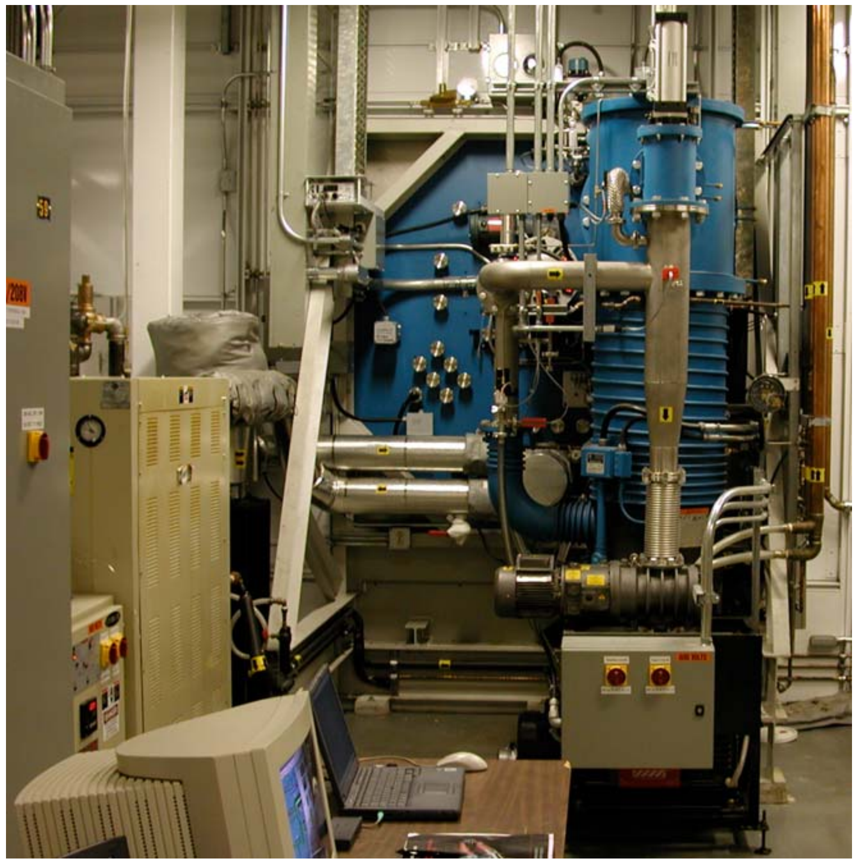

Mammography Detector

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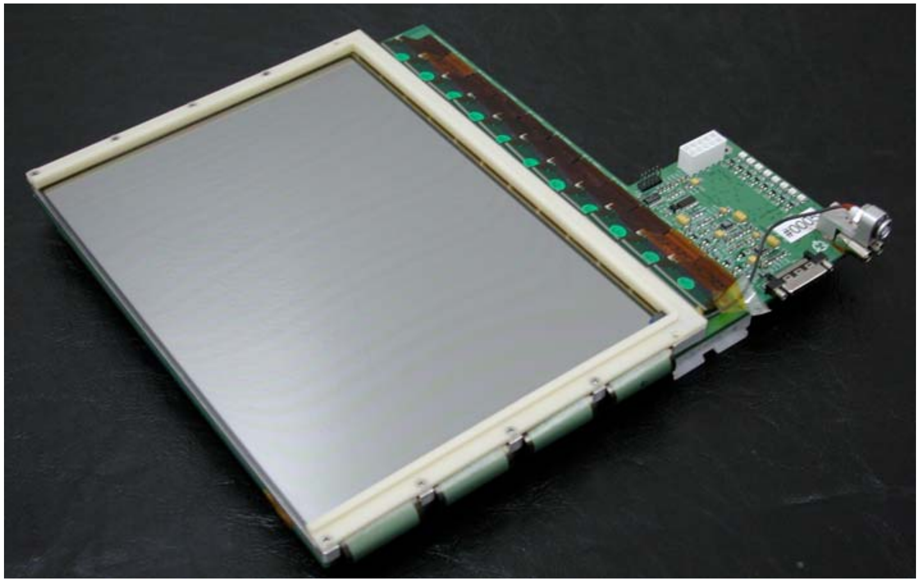

Electronic sub-assembly

Packaged detector

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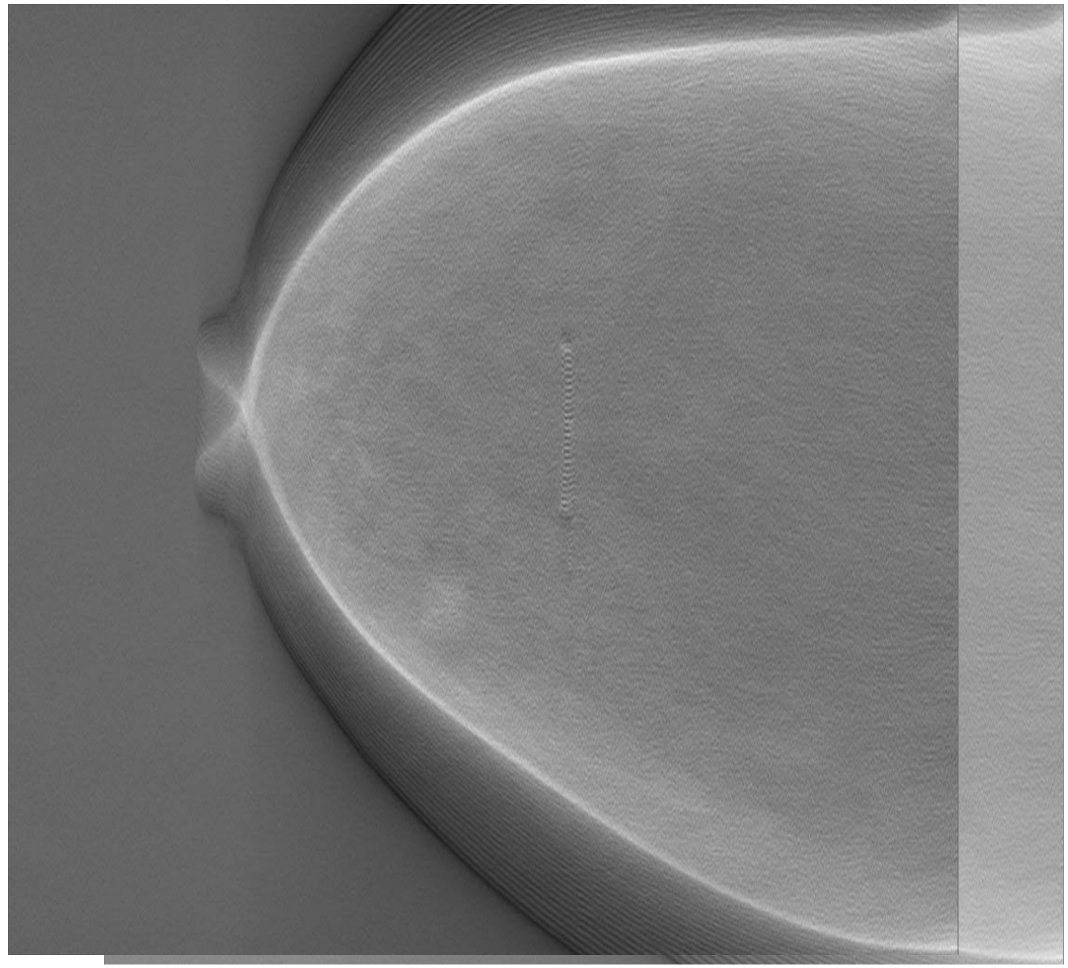
Tomosynthesis



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