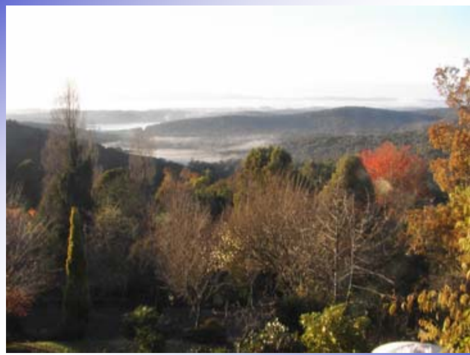
Imaging with Synchrotron Light

Rob Lewis
Monash University
Australia

Basics of X-ray Imaging

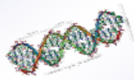


General Radiography

- 2-D image from 3-D object (patient)
- Map of attenuation for each photon path
- Essentially an image of density, or structure

Hierarchy of Biological Organisation

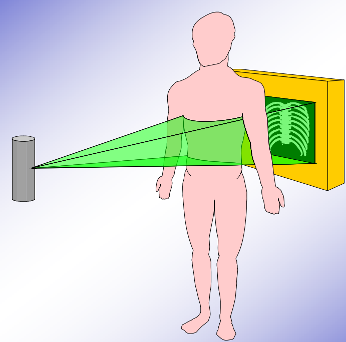
Organism	m
Organ System	m
Organ	cm
Tissue	mm
Cell	10µm
Organelle	1µm
Macromolecules	100nm
Molecules	0.1nm



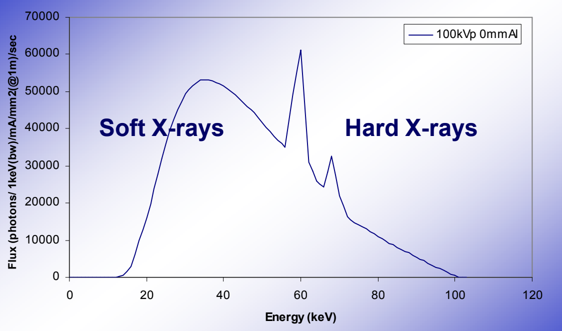
Imaging Requirements

- High spatial resolution
  - Preferably sub cellular (µm)
- High temporal resolution
  - Cell kinetics, Chemokinetics (ms to µs)
- High anatomical contrast
- Specificity for tracers

Conventional Radiography



X-ray Spectra



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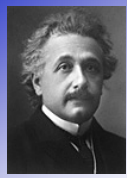
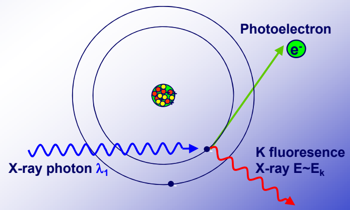
Albert Einstein



Germany and Switzerland
Kaiser-Wilhelm-Institut
(now Max-Planck-Institut)
für Physik
Berlin-Dahlem, Germany
1879 - 1955

Nobel prize in physics 1921
"for his services to Theoretical Physics, and especially for his discovery of the law of the photoelectric effect"

Photoelectric Effect



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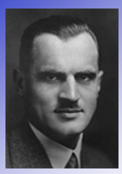
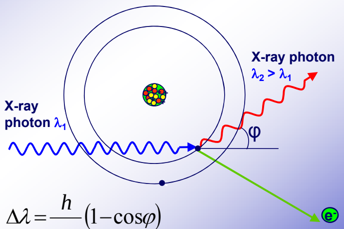
Arthur Holly Compton



University of Chicago
Chicago, IL, USA
1892 - 1962

Nobel prize in physics 1927
"for his discovery of the effect named after him"

Compton Effect



$$\Delta\lambda = \frac{h}{m_0 c}(1-\cos\varphi)$$

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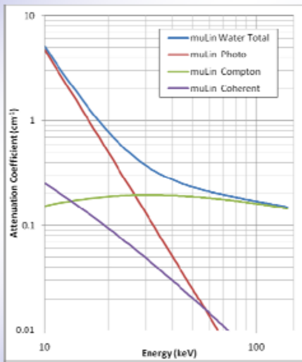
Water Attenuation Coefficients

- Photo electric falls rapidly with increasing energy
- Compton scattering roughly constant
- Coherent scattering falls with increasing energy but less rapidly than photoelectric (important see later)



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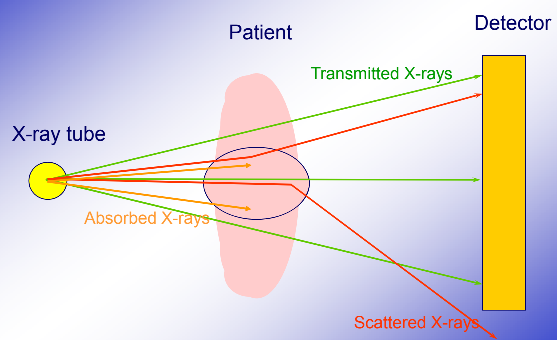
Radiographic Imaging



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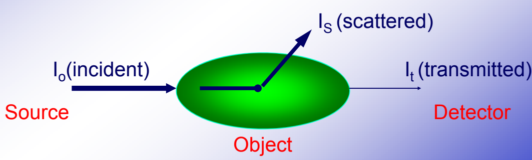
Interaction of X-rays & Matter

- Interaction with matter (human tissue)
  - Scattering (loss of energy & direction change)
  - Absorption (energy transferred to tissue)
- Scattering + absorption = attenuation



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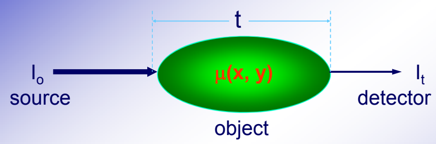
X-ray Attenuation



- μ = linear attenuation coefficient (cm^{-1})
- Homogeneous: $I_t = I_o \exp -[\mu t]$
- Inhomogeneous: $I_t = I_o \exp -[\Sigma\mu\, dt]$
- $\ln (I_o/ I_t) = \log_e (I_o/ I_t) = \mu t$ = attenuation/ray-sum

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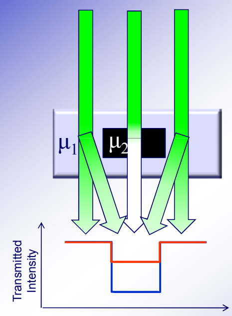
X-rays and Contrast



- Difference in attenuation coefficients generates contrast
- $\mu_1 < \mu_2$
- Scatter reduces contrast

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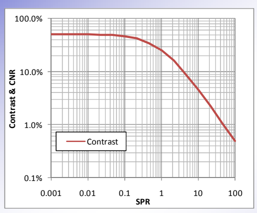
Scatter in Medical Imaging



Examination	Energy	Field Size (cm)	Antiscatter mechanism	$Scatter_{Total}$/ Primary	$Scatter_{Coherent}$ / $Scatter_{Total}$
Chest	120 kVp	30 x 30	6.7 cm air gap	2.3	0.12
			20 cm air gap	1.2	0.11
Abdomen	80 kVp	17 x 17		2.7	0.26
			Grid	0.34	0.075
Mammography	30 kVp (Mo)	12 cm diam	Grid	0.6	0.24

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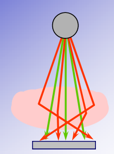
Air Gap



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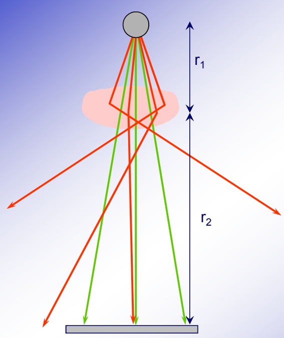
Air Gap



- Air gap is good because
  - No mechanics
  - No increase in dose
- But
  - Increases the effect of penumbral blurring
  - Blurring=f r_2/r_1 where f is the size of the focal spot
  - For f = 0.5mm & r_1=1m blurring= $5\times10^{-4}\, r_2$
  - This translates into a lateral coherence of 0.1μm

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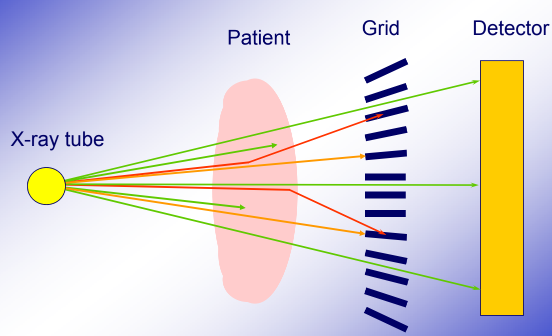
Use of Grid to Remove Scatter



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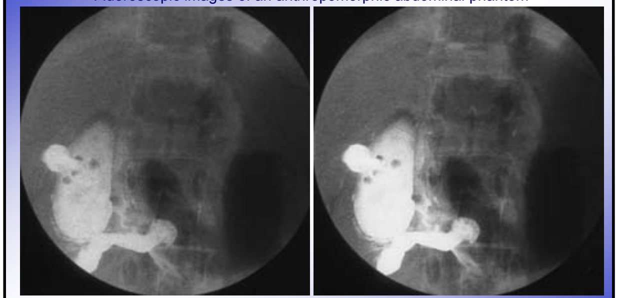
Effect of Antiscatter Grid.

Fluoroscopic images of an anthropomorphic abdominal phantom



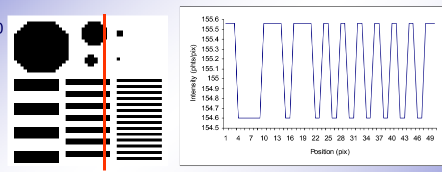
With Grid | Without Grid

JP Taubel, B A Schueler, TJ. Vrieze, JE. Gray *Radiographics.* 2001;21:507-518

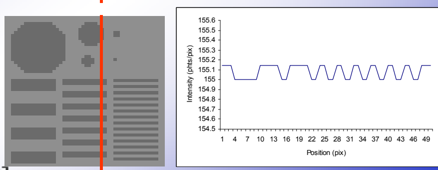
Contrast

	Surround	Object
Bone	0.5cm	0.5cm
Brain	12cm	10cm
Soft tissue	0cm	2cm

Scatter/Primary=0



Scatter/Primary=3



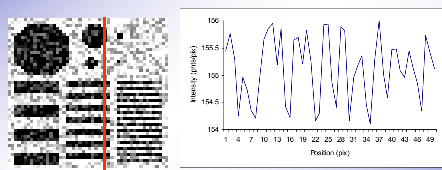
Use of grids

- Grids reduce x-ray scatter
- Grids improve film contrast & resolution
- Grids increase patient dose

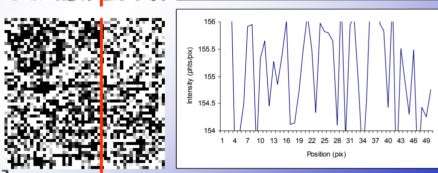
Noise

	Surround	Object
Bone	0.5cm	0.5cm
Brain	12cm	10cm
Soft tissue	0cm	2cm

Noise=1



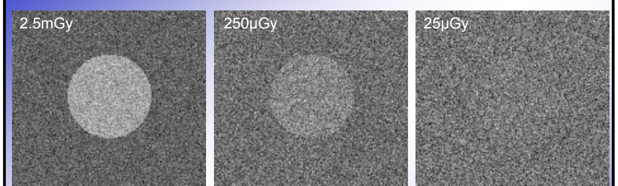
Noise=3



Counting Statistics

- Photons are quantised and hence subject to probabilities
- The Poisson distribution expresses the probability of a number of events occurring in a time period
- If the expected number is n then $P(n,k) = \frac{n^k e^{-n}}{k!}$
- The mean of P(n,k) is n
- The variance of P(n,k) is n
- The standard deviation is √n
- Fractional error = (√n)/n=1/√n
- As n increases, uncertainty and noise decrease

Photon Noise (Quantum Mottle)

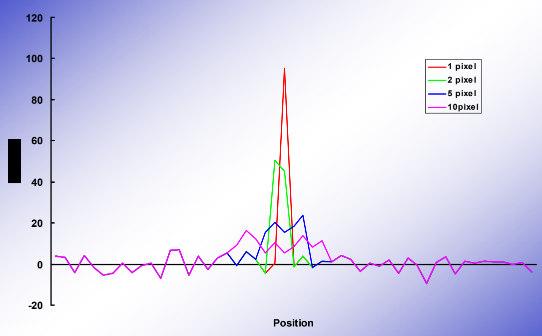


Simulated images of a 1mm thick Aluminium disc in 20cm of water recorded by a perfect detector

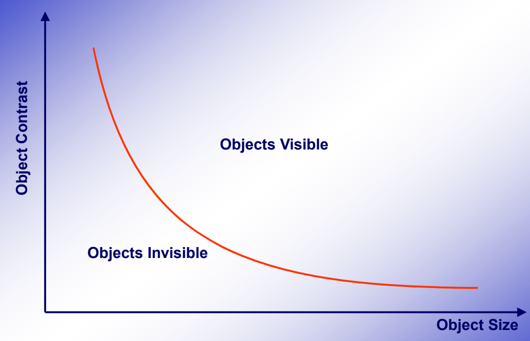
Illumination = Tungsten anode operated at 120kVp with 3mm Al filtration

Entrance doses as shown

Effect of Resolution



Contrast Detail Curve



Contrast, Resolution & Noise

- Noise degrades image
  - Arises from many things
    - No of photons, i.e. source
    - Scatter, age, processing
- Resolution
  - Specifies how sharp an image is
  - Specified as
    - As distance between features, d (mm)
    - As LPPM (line pairs per mm), d^{-1} (mm^{-1})
    - As MTF, in LPPM at given percentage MTF
- Contrast - ability to differentiate regions
  - Grey scale regions G_A and G_B
  - Contrast, $C = 100*(G_A - G_B)/(G_A + G_B)$ %

Radiography and Dose

- Transmitted X-rays deposit no dose
- Absorbed X-rays deposit dose
- Scattered X-rays also deposit dose.

X-ray Dose

- Exposure
  - Quantity of X-rays as measured by ionisation in air is exposure
  - Kinetic Energy Released in Matter (KERMA)
  - Units are C kg^{-1}
  - Old unit is the Roentgen (R) ($1R=2.58\times10\text{-}4$ C kg^{-1})
- Dose
  - Energy absorbed
  - Units are J kg^{-1} = Gy   Old unit is the rad. (1 rad =0.01Gy)
- Equivalent dose
  - Not all types of radiation cause the same biological damage per unit dose
  - Unit is the Sievert   Old unit is the rem (1 rem=0.01Sv)
  - Equivalent dose $H(Sv)=w_R(Sv)\times D(Gy)$
  - For X-rays $w_R=1$

How Does Radiation Interact with the Body?

- When ionising radiation is absorbed in the body it can create free-radicals which may alter normal cell function
  - E.g. DNA damage leading to genetic mutations
  - At high doses (> 1 sievert) this can result in massive cell death, organ damage and death to the individual.
  - At low doses (less than 50 mSv) the situation is more complex
    - Nb. DNA repair mechanisms
- A dose of one millisievert
  - Individual
    - A chance of 6 in 100,000 of contracting cancer
    - Compare with normal lifetime cancer incidence of 25,000 cases per 100,000
  - A large population
    - Will produce two cases of severe hereditary effects per million births
    - Compare with the normal incidence of severe congenital abnormalities which is 23,000 per million births

Man's Exposure To Ionizing Radiation

Source Of Exposure	Exposure
Natural Radiation (Terrestrial and Airborne)	1.2 mSv per year
Natural Radiation (Cosmic radiation at sea level)	0.3 mSv per year
Total Natural Radiation	**1.5 mSv per year**

Seven Hour Aeroplane Flight	0.05 mSv
Chest X-Ray	0.04 mSv
Nuclear Fallout (From atmospheric tests in 50's & 60's)	0.02 mSv per Year
Chernobyl (People living in Control Zones near Chernobyl)	10 mSv per year
Cosmic Radiation Exposure of Domestic Airline Pilot	2 mSv per year

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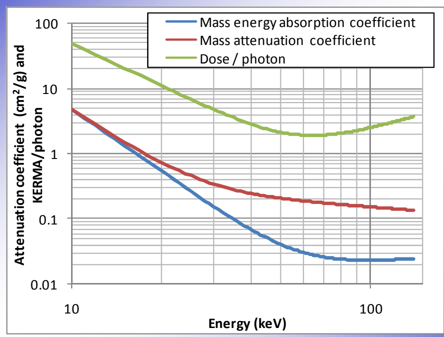
http://www.arpansa.gov.au/is_rad.htm

Dose Deposited in Soft Tissue



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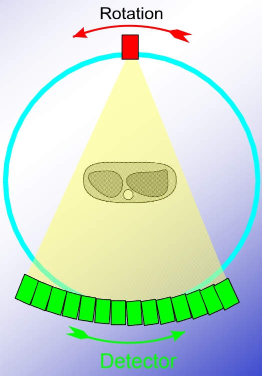
3rd Generation CT Scanner

- Multiple detectors
- Translation-rotation
- Large fan beam
- Patient stationary for each 2-D slice acquisition; about 1-2 seconds per slice
- kV = 120, mA = 500
- Image then reconstructed in about 1-2 seconds



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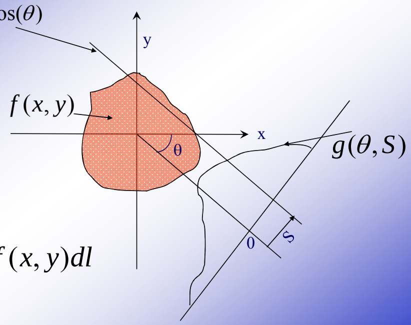
Projection and the Radon transform

Equation of this line l is:
$S = x\sin(\theta) - y\cos(\theta)$

$$g(\theta, S) = \int_l f(x,y)dl$$



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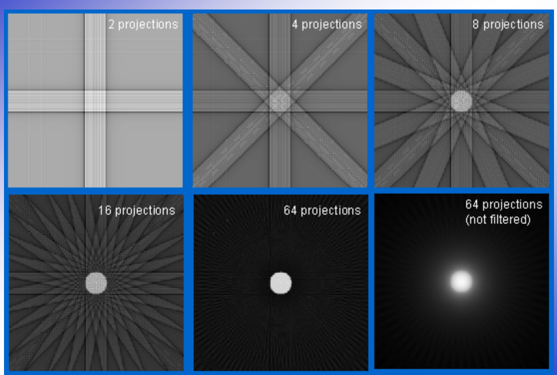
FBP in Practice



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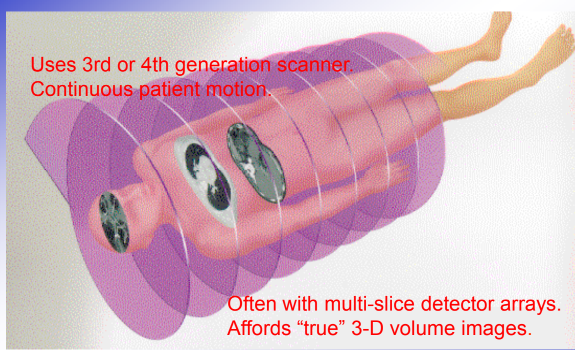
Volume CT image



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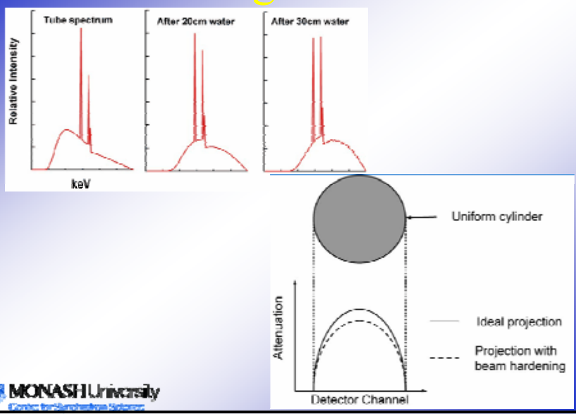
Beam Harding Artefacts



Beam Harding Artefacts

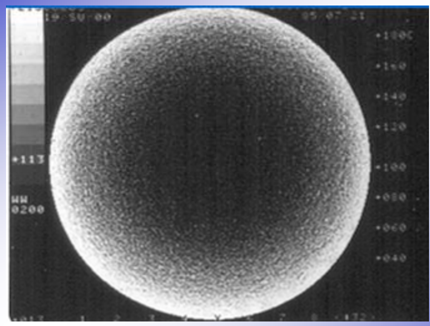
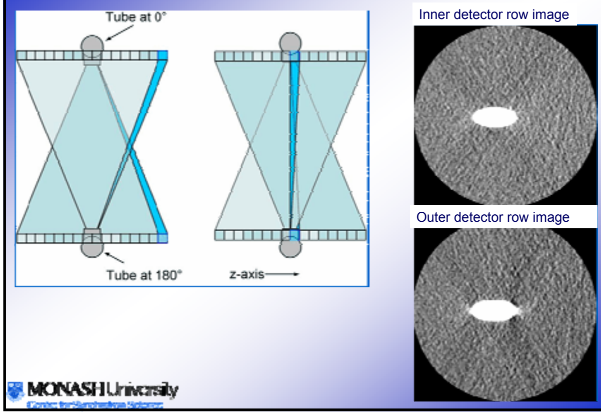


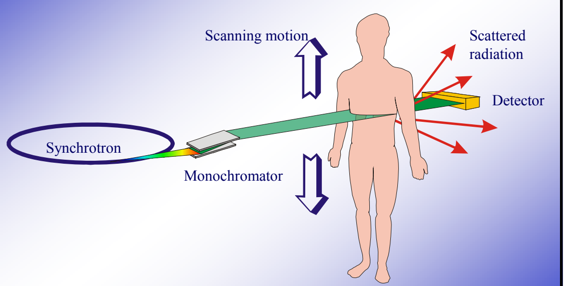
Image of uniform phantom

Cone Beam Artefacts

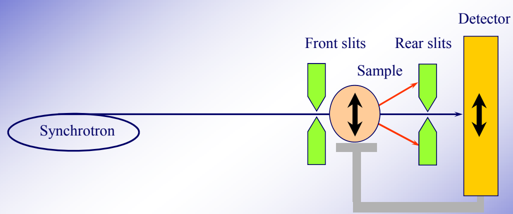


Synchrotron Differences

Synchrotron Radiography



SR Radiography



Synchrotron

<u>Advantages</u>

- Monochromatic
  - No beam hardening artefacts
- Parallel beam
  - No cone beam artefacts
  - Reduced scatter
- High Intensity
  - High speed

<u>Disadvantages</u>

- Source cannot be rotated
  - Object must be rotated which limits speed of CT
- Beam height is limited, especially at high energies
  - Object may have to be scanned
  - Limits size of object and/or time resolution

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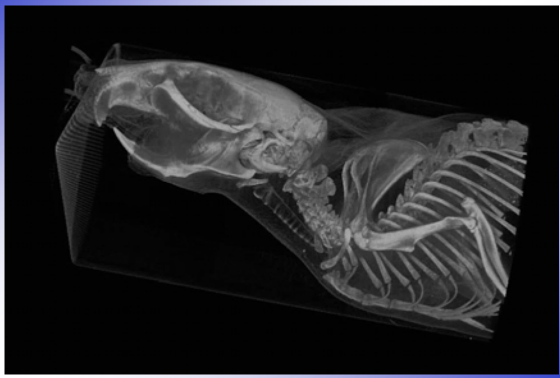
Mouse CT



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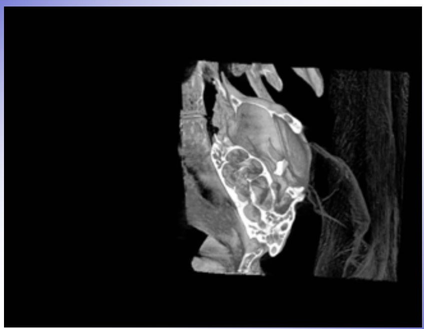
David Parsons and Karen Siu

Mouse Cochlea



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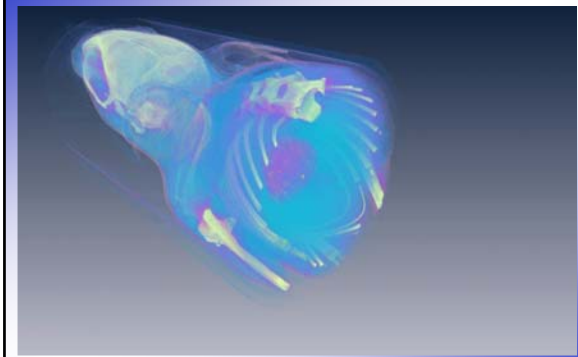
David Parsons and Karen Siu

Mouse Fly Through



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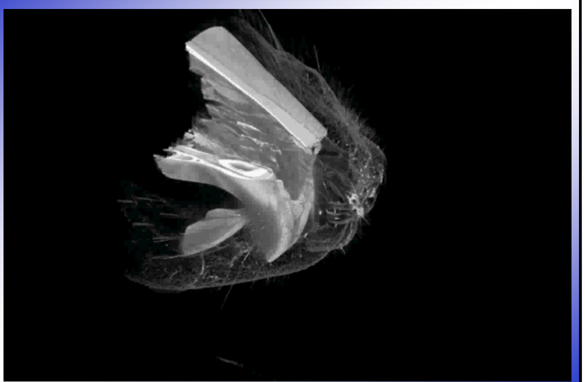
Kaye Morgan

Mouse CT



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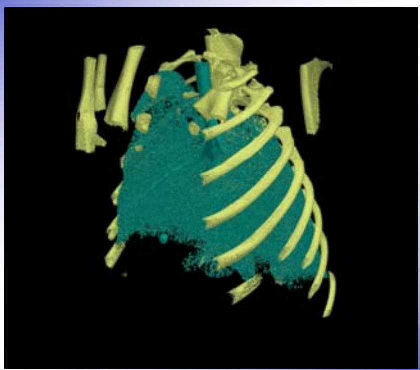
David Parsons and Karen Siu

Rabbit Pup CT



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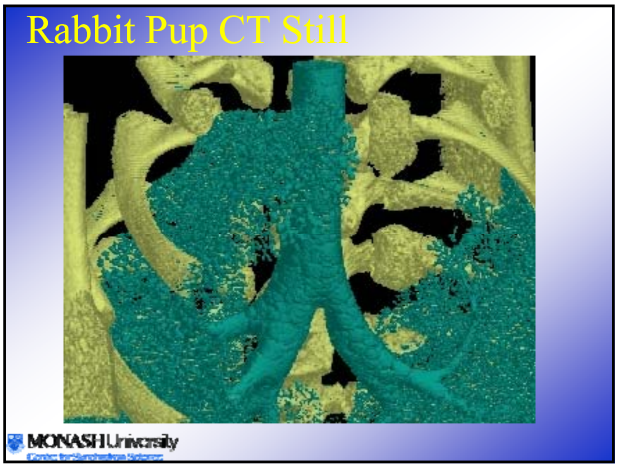
Rabbit Pup CT Still
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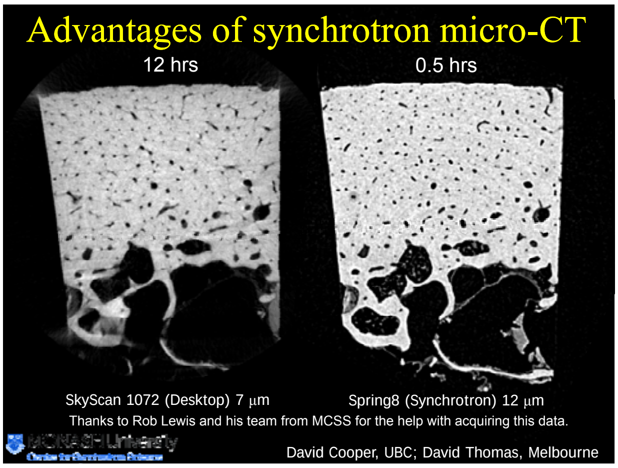
Advantages of synchrotron micro-CT
12 hrs
0.5 hrs
SkyScan 1072 (Desktop) 7 μm
Spring8 (Synchrotron) 12 μm
Thanks to Rob Lewis and his team from MCSS for the help with acquiring this data.
David Cooper, UBC; David Thomas, Melbourne


The Search for Contrast
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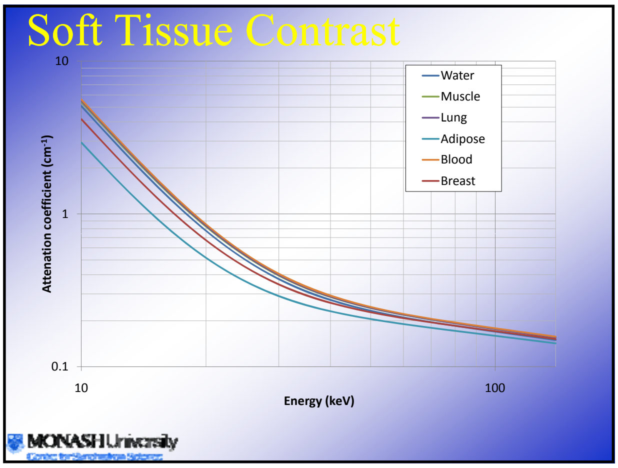
Soft Tissue Contrast
Attenuation coefficient (cm⁻¹)
10
1
0.1
10
100
Energy (keV)
Water
Muscle
Lung
Adipose
Blood
Breast
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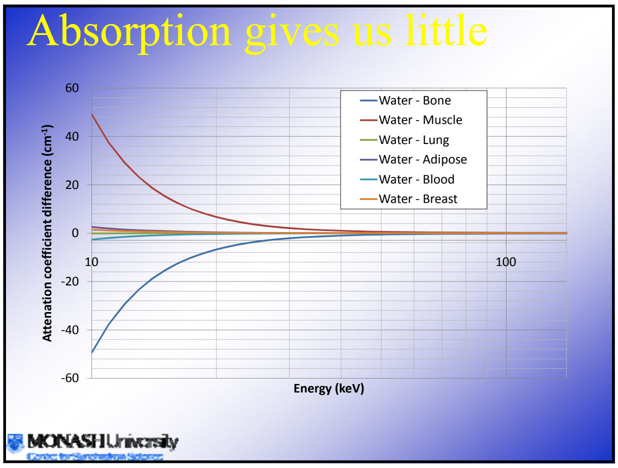
Absorption gives us little
Attenuation coefficient difference (cm⁻¹)
60
40
20
0
-20
-40
-60
10
100
Energy (keV)
Water - Bone
Water - Muscle
Water - Lung
Water - Adipose
Water - Blood
Water - Breast
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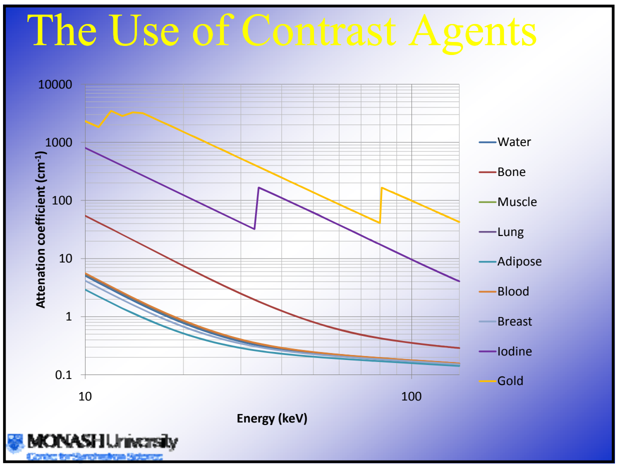
The Use of Contrast Agents
Attenuation coefficient (cm⁻¹)
10000
1000
100
10
1
0.1
10
100
Energy (keV)
Water
Bone
Muscle
Lung
Adipose
Blood
Breast
Iodine
Gold
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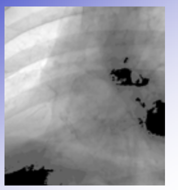
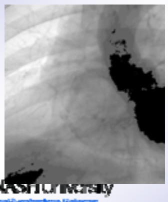
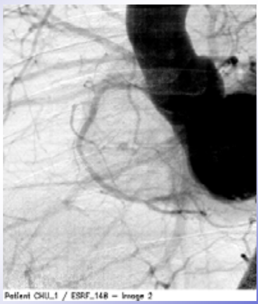
Dual Energy Digital Subtraction

High



Low





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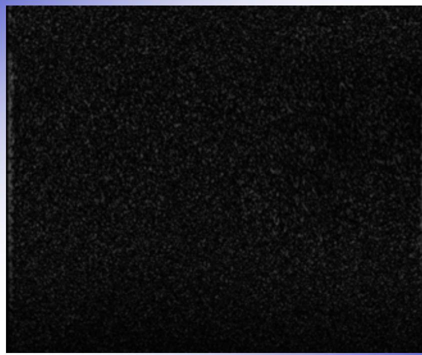
W. Thomlinson

Kidney Movie (Mouse)



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James Pearson, Physiology

Pulmonary Hypertension (Mouse)

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James Pearson, Physiology

Tracking Cells in-vivo

- C6 glioma cells labeled with gold nano-particles
- Cell implantation
  - small volume injected over extended time period
  - minimal mechanical injury from

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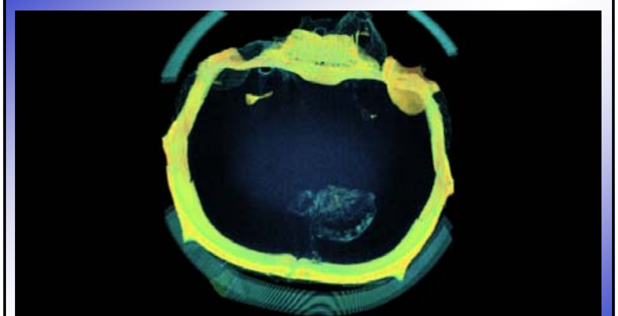
U. Saskatchewan, Sincrotrone Trieste , Monash University, Charles Gardiner Hosp.

Rat 3: 600000 cells, 14 days incubation



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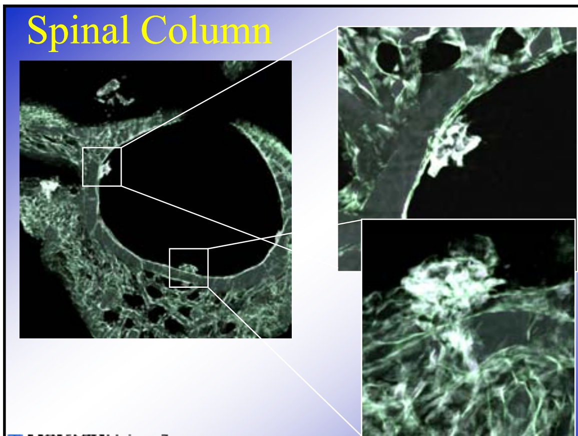
U. Saskatchewan, Sincrotrone Trieste , Monash University, Charles Gardiner Hosp.

Spinal Column



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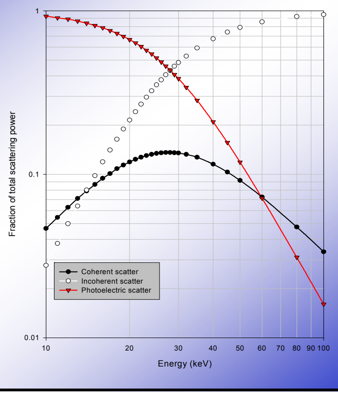
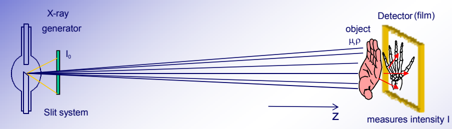
Scatter processes in biological tissue

Fraction of total scattering power as a function of x-ray energy



Radiographic imaging



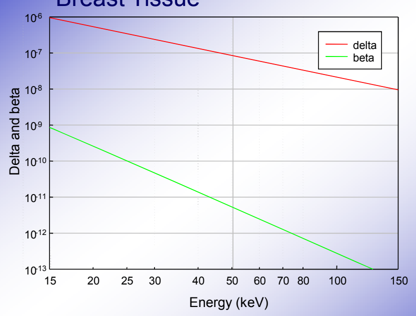
$$I(x,y)=I_0\cdot e^{-\int_{-z_0}^{z_0}\frac{\mu(x,y,z)}{\rho(x,y,z)}\cdot\rho(x,y,z)dz} = E^{*}E$$

$$E=E_0\cdot e^{-i\cdot x\cdot\left(\frac{2\pi}{\lambda}\underbrace{-n_a\cdot r_0\cdot\lambda\cdot f_1}_{\text{phase}}+\omega t\underbrace{+i\cdot n_a\cdot r_0\cdot\lambda\cdot f_2}_{\mu}\right)}$$

Phase shift ~ f_1
scattering angle ~ ∇f_1

Ralf Menk, Elettra

Complex Refractive Index

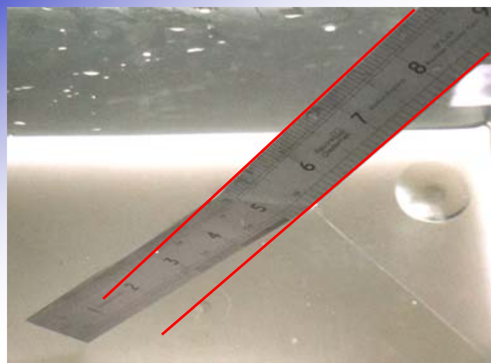


Refractive index
η = 1- δ -iβ

Where β = absorption
δ = phase shift

Nb.
δ ~ 1000 β
δ ~ E^{-2}
β ~ E^{-4}

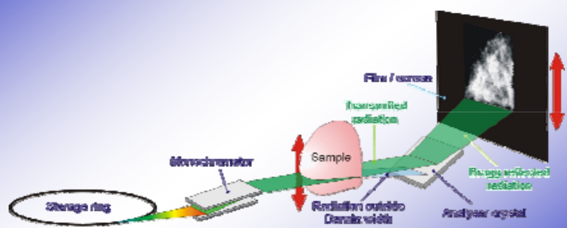
Refraction



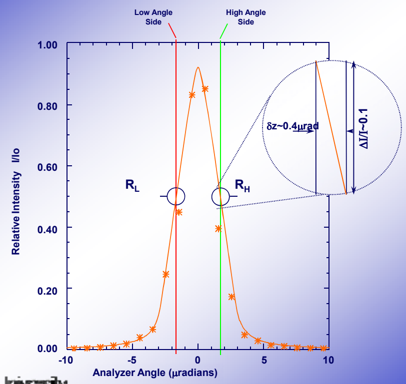
Phase Contrast 1: ABI

Analyser Based Imaging

Sometimes called Diffraction Enhanced Imaging



Crystal Rocking Curve



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Beamline Camera

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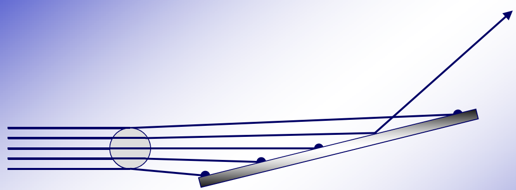
Rocking Curve



Refractive index for X-rays is less than 1 by about 1 part in a million

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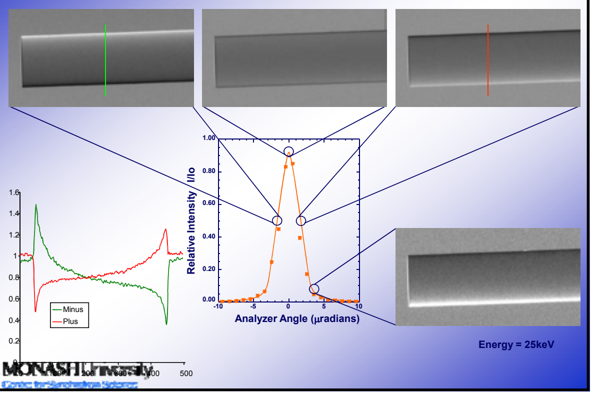
ABI How it works



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ABI Mathematics

- I_L & I_H = Intensities on low and high angle sides of rocking curve
- $Grad_L$ & $Grad_H$ = Gradients of low and high angle sides of rocking curve
- I_R is intensity
- $\Delta\theta_z$ = refraction angle

Given

$$I_L = I_R \cdot \left(R_L + Grad_L \cdot \Delta\theta_Z\right)$$

$$I_H = I_R \cdot \left(R_H + Grad_H \cdot \Delta\theta_Z\right)$$

$$\mathrm{Find}\left(I_R, \Delta\theta_Z\right) \rightarrow \begin{pmatrix} \dfrac{Grad_H \cdot I_L - Grad_L \cdot I_H}{Grad_H \cdot R_L - Grad_L \cdot R_H} \\ \dfrac{I_H \cdot R_L - I_L \cdot R_H}{Grad_H \cdot I_L - Grad_L \cdot I_H} \end{pmatrix}$$

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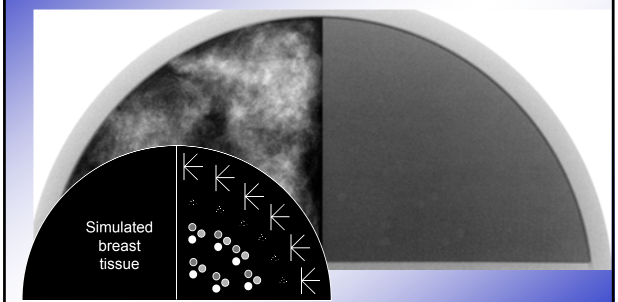
TORMam Conventional



Spectrum = Mo:Mo 28kVp

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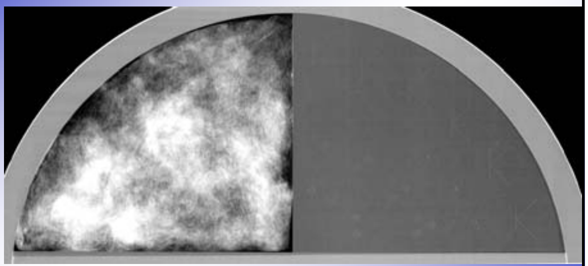
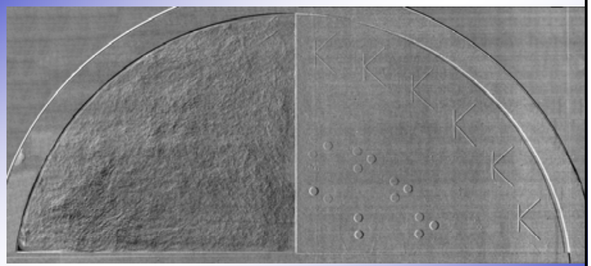
TORMAM Peak



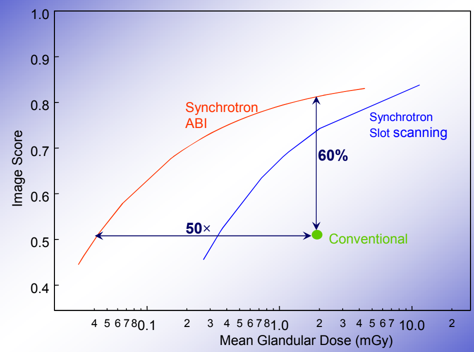
Energy = 20keV

TORMAM Refraction



Energy = 20keV

ABI vs Slot Scanning



Human Finger

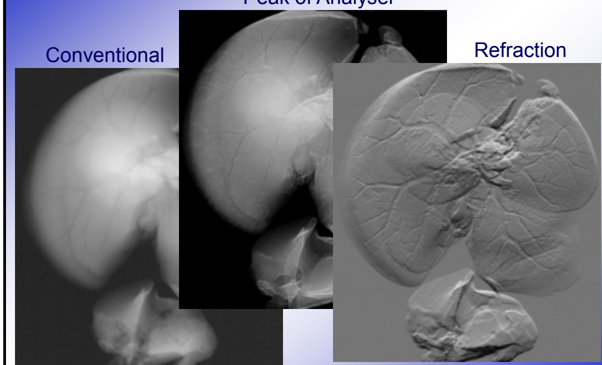
Conventional Peak of Analyser Refraction

Liver

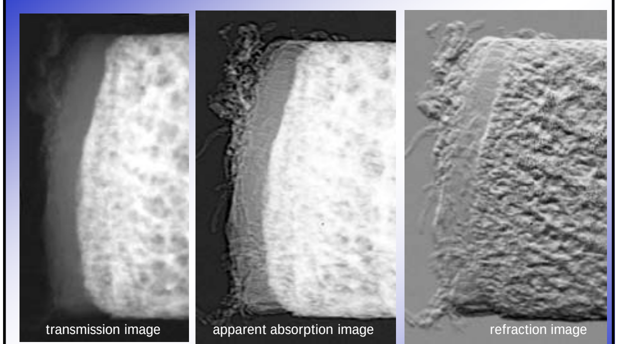
Peak of Analyser

Conventional

Refraction



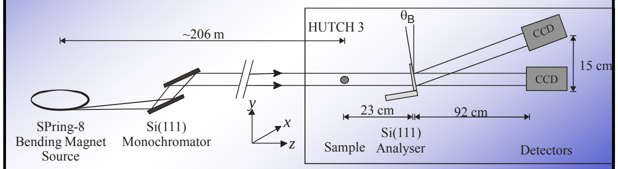
Femur head core cuts



Sharmila Majumdar and Ahi Sema Issever , University of California, Luigi Rigon Univ. Trieste

Imaging Setup

- Beamline 20B2, Spring-8, Japan.
- 25 keV beam of highly coherent radiation.



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Marcus Kitchen, Monash University

Laue analyser crystal

- 100 µm thick Si(1,-1,0) surface-cut designed for 25 keV X-ray beam

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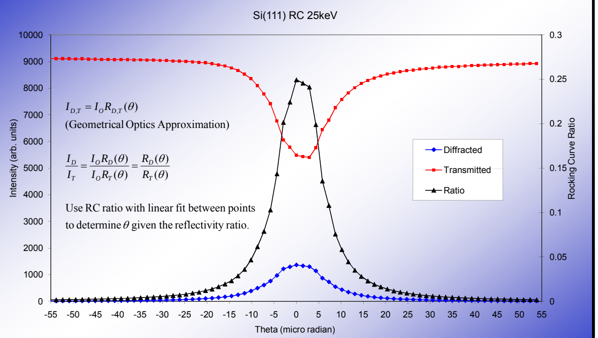
Rocking Curves and their Ratio



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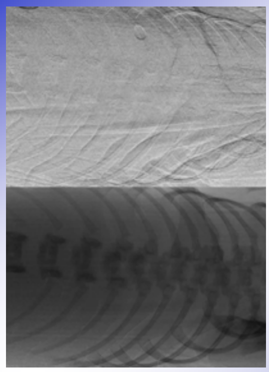
Laue based ABI of Rabbit



- Refraction angle image
- Apparent absorption image
- Exp=0.3 s, Int=2 s, 20 fps

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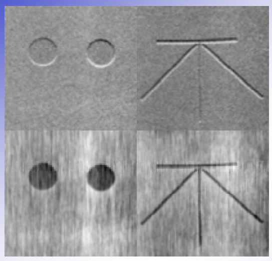
From Phase to Absorption



- Back to the phase, and electron density

$$I_{DEI}(X,Y) = I_{IN}(X,Y)\left[R(-\xi_0) + R'(\xi-\xi_0)\frac{\partial}{\partial X}\varphi_{IN}(X,Y) - \alpha\frac{\partial}{\partial X}\log_e I_{IN}(X,Y)\right]$$

$$\varphi_{IN}(X,Y) = \int dX\left[-\mathrm{Re}\left(\frac{S}{T}\right) + \mathrm{sgn}\left(\mathrm{Re}\left(\frac{S}{T}\right)\right)\sqrt{\left(\mathrm{Im}\left(\frac{S}{T}\right)\right)^2 + \frac{I_{OUT}(X,Y)}{I_0|T|^2}}\right]$$

D. Paganin, T.E. Gureyev, K.M. Pavlov, R. A. Lewis and M. Kitchen, "Phase retrieval using coherent imaging systems with linear propagators", Optics Communications, 234(2004) 87-105

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Phase Contrast 2: Interferometry

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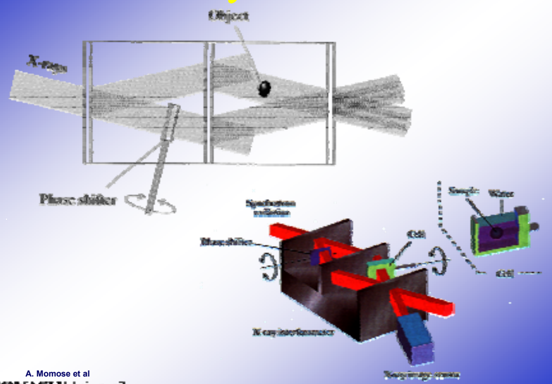
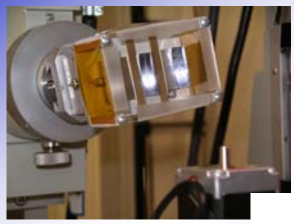
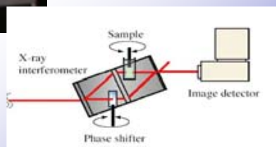
Interferometry



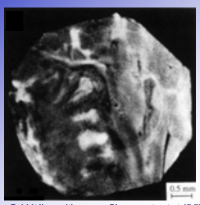
A. Momose et al

An x-ray interferometer



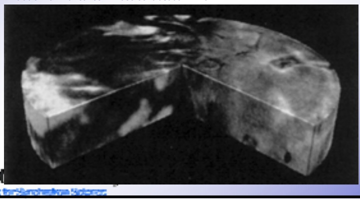


Interferometry



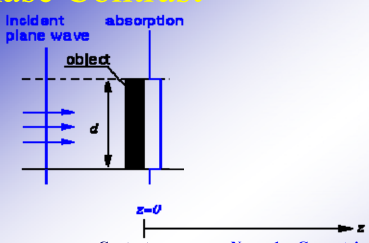
Histopathology slice

Rabbit liver with cancer. Phase contrast at 17.7keV



Phase Contrast 3: Propogation BI

Phase Contrast

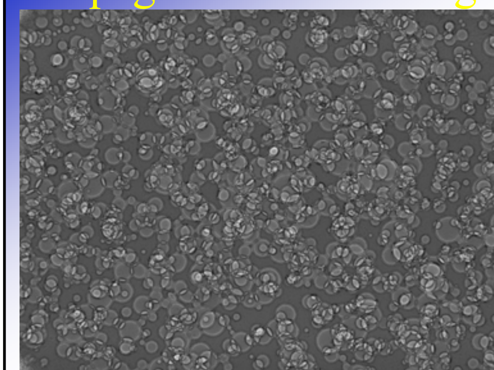


$$N_F = \frac{d^2}{\lambda z}$$

- **Contact:** $N_F >> 1$ **Geometric approximation**
  - The intensity distribution is a pure absorption image.
- **Near field:** $N_F >> 1$ **Geometric approximation**
  - Contrast is given by sharp changes in the refractive index, i. e. at interfaces.
- **Intermediate field:** $N_F \sim 1$ **Fresnel approximation**
  - The image loses more and more resemblance with the object.
- **Far field:** $N_F << 1$ **Far: Fraunhöfer approximation**
  - The image is the Fourier transform of the object transmission function

Timm Weitkamp ESRF ID22

Propagation Based Imaging



147cm

Using SR light characteristics

- Coherence
  - The coherence length is less than a micron on older sources but at third generation sources gets up to 100's of microns.
  - This opens up potentials for several coherence imaging techniques.

Phase contrast imaging techniques

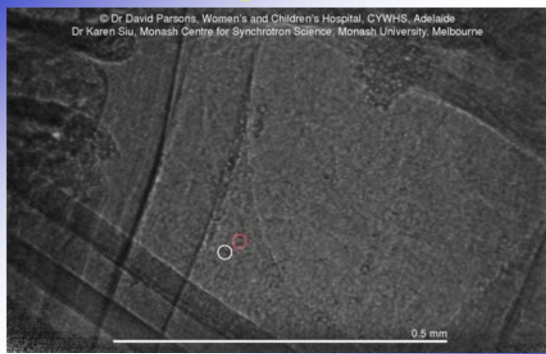
So far:

- Interferometry – Phase directly
- Analyser based imaging – Gradient of the phase (differential $\nabla\phi$)
- Propagation based imaging – Laplacian of the phase (second order differential $\nabla^2\phi$).

Mouse and Nebuliser

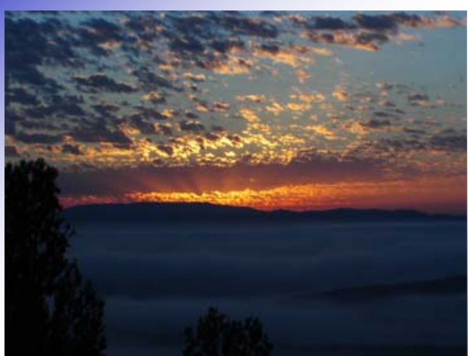
David Parsons and Karen Siu

Bead Tracking

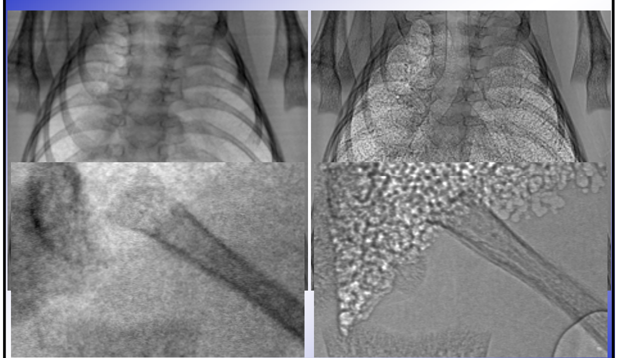


David Parsons and Karen Siu

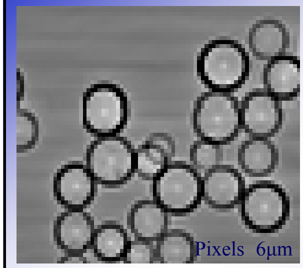
A Case Study: Lung Imaging

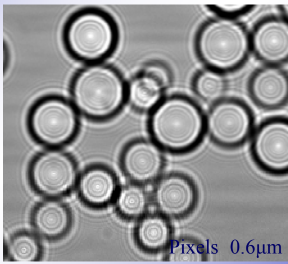


Rabbit Lung



Glass Spheres BL20XU

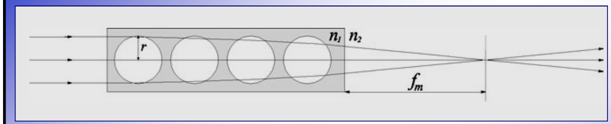




Energy	20 keV
R_2	1.45m
Spheres	45-75µm

N. Yagi, K. Uesugi, SPring-8

Refractive lenses?

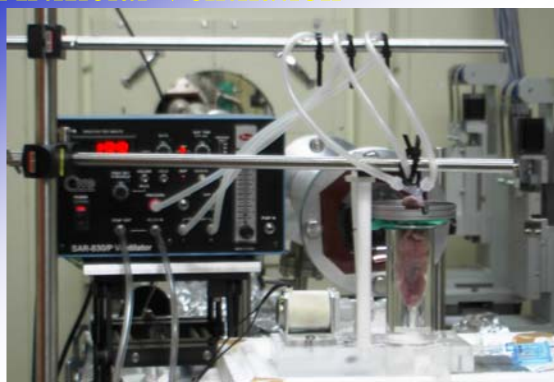


- 10 mm thick lung: Alveoli ~ 60 μm ø
  - Max number of aligned spheres is ~ 150
- At 33 keV, assuming ~ 50% aligned
  - Max number of aligned spheres is ~ 75
- Consequently f_m ~ 0.7 m.

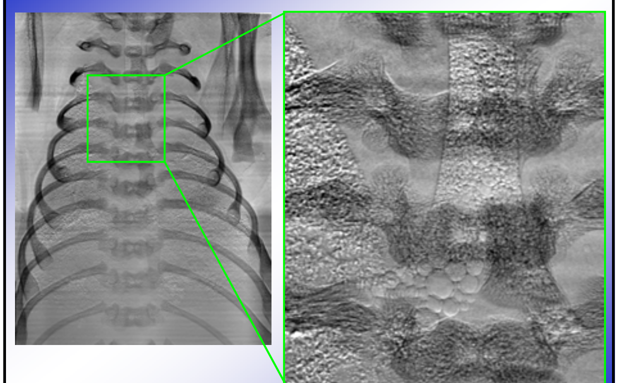
Birth:
One of the greatest Physiological challenges

- During fetal life the future airways of the lungs are liquid-filled
- At birth lungs must rapidly transform from being liquid to air filled
- How this happens is poorly understood but the process
  - Develops late in pregnancy
  - Is initiated by labour
- Preterm and caesarean section infants can develop problems
- Lack of clearance
  - Requires weeks of assisted ventilation (>$2,000/day)
  - ~30% develop chronic lung disease
  - Incidence is increasing
- Our ability to study the problem has been greatly inhibited by the lack of a suitable imaging modality

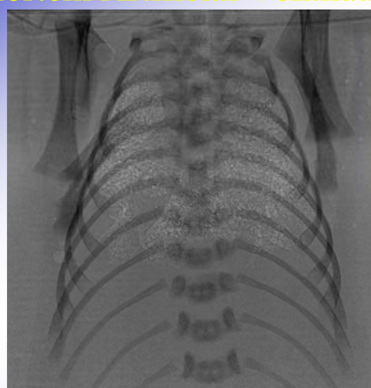
Artificial Ventilation



Surfactant



Post Mortem Artificial Ventilation



Rabbit Pup Post Mortem Pneumothorax

Phase to Projected Thickness

- Can we retrieve quantitative information from these images?
- Full phase retrieval
  - If no assumptions are made 4 images required
- Single image phase retrieval assumptions
  - Plane waves
  - Single-material object embedded in a substrate of approximately uniform thickness
- Lungs (air) in the chest (water)

David Paganin, Timur E. Gureyev, Konstantin M. Pavlov, Robert A. Lewis, Marcus Kitchen; Optics Communications 234 (2004) 87–105

Phase Retrieval: Single Image

- Approximate 'contact' intensity from Beer's Law

$$I(\mathbf{r}_\perp, z=0) = I_O \exp(-\mu T(\mathbf{r}_\perp))$$

- Approximate 'contact' phase by

$$\phi(\mathbf{r}_\perp, z=0) = -\frac{2\pi}{\lambda}\delta T(\mathbf{r}_\perp)$$

- Use Transport-of-Intensity Equation (TIE)

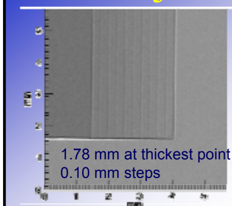
$$\nabla_\perp \cdot (I(\mathbf{r}_\perp, z)\nabla_\perp \phi(\mathbf{r}_\perp, z)) = -\frac{2\pi}{\lambda}\frac{\partial}{\partial z} I(\mathbf{r}_\perp, z)$$

- Solve for object's projected thickness using Fourier Derivative Theorem

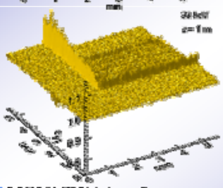
$$T(\mathbf{r}_\perp) = -\frac{1}{\mu}\ln\left(\mathbf{F}^{-1}\left\{\mu\frac{\mathbf{F}\{M^2 I(M\mathbf{r}_\perp, z=R_2)\}/I_O}{MR_2\delta|\mathbf{k}_\perp|^2 + \mu}\right\}\right)$$

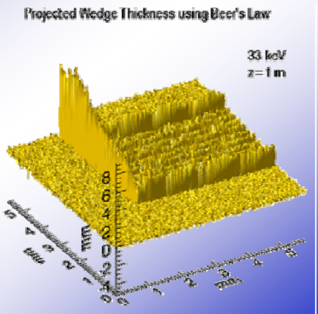
Paganin, D et al., Journal of Microscopy, Vol. 206 Pt 1 April 2002, pp. 33

Test Object – Perspex Step Wedge



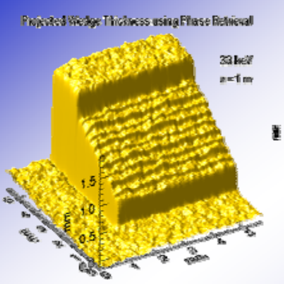
Noise: 1 std dev = 0.5% of mean

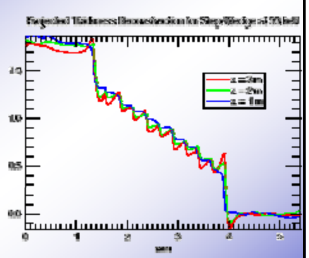




Marcus Kitchen, Monash

Phase Retrieval



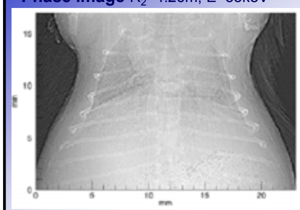


- Step height matches measured thickness
- Noise, surface defects and sharp diffraction edges produce roughness in the projected thickness.
- μ and δ must be well known.

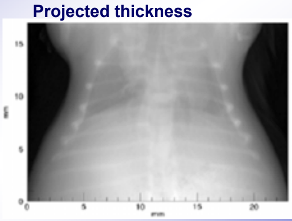
Marcus Kitchen, Monash

Phase to Projected Thickness

Phase image R_2=4.26m, E=33keV



Projected thickness

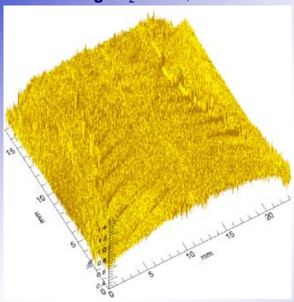


- Surface Entrance Dose ~ 8.6 mGy
- 1% Noise
- Use μ and δ for lung tissue at 33 keV
- Projected thickness resembles conventional radiograph

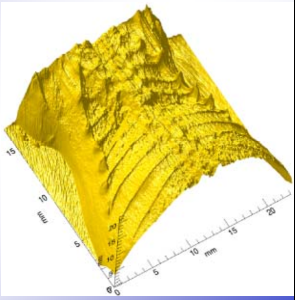
Marcus Kitchen, Monash

Phase to Projected Thickness

Phase image R_2=4.26m, E=33keV



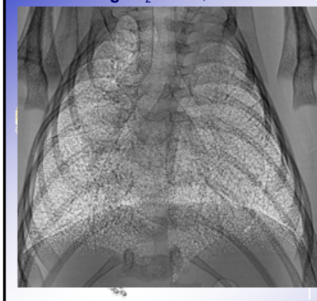
Projected thickness



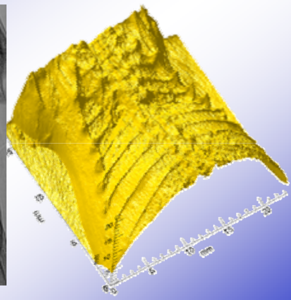
Marcus Kitchen, Monash

Phase to Projected Thickness

Phase image R_2=4.26m, E=33keV



Projected thickness



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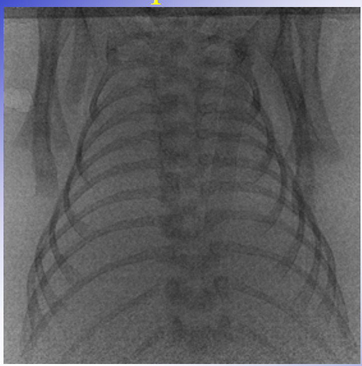
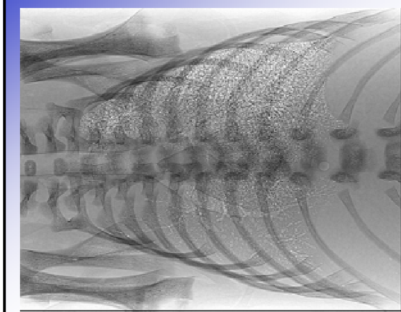
Rabbit Spontaneous Breathing



- Major inspiratory activity important for aeration

RA Lewis et al Phys. Med. Biol. **50,** 5031-5040
S. Hooper et al FASEB **21**, 3330 (2007)

Non-uniform Aeration

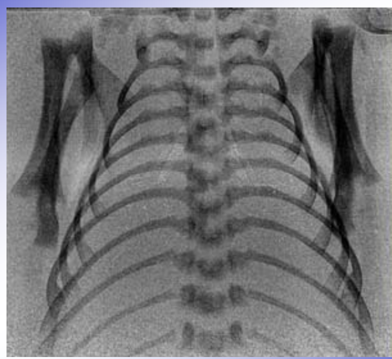


Dependent lung aerates less well than non-dependant lung

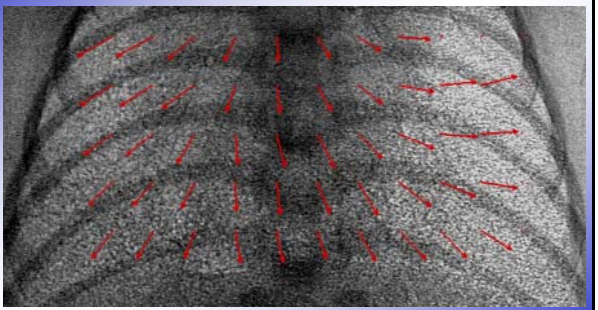
Hooper et al FASEB **21**, 3330 (2007)

Rabbit Pup: No PEEP

Rabbit Pup: With PEEP



Detecting lung shear stress during ventilation "Particle image Velocimetry"



A. Fouras, J. Dusting, R. Lewis, and K. Hourigan J. Appl. Phys. 102, 064916, 2007

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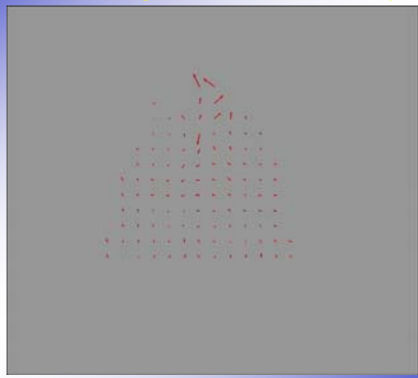
Particle image velocimetry



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A. Fouras, J. Dusting, R. Lewis, and K. Hourigan J. Appl. Phys. 102,

Blood Flow. No Contrast Medium

570μm

190nm pixels

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A. Fouras, J. Dusting, R. Lewis, and K. Hourigan, J. Appl. Phys. 102, 064916, 2007
B. Sally Irvine et al Applied Physics Letters (accepted)

Phase Contrast Comparison

	Interferometer	Analyser	Propagation
Requirements			
Spatial coherence	High	Reasonable	High
Temporal coherence	High	High	Low
Capabilities			
Large objects	Very difficult	Difficult	Easy
Movies	Hard	Hard	Easy
Sensitivity	Outstanding	Excellent but unidirectional	Good

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How Many Does it Take?

- Monash Physics
  - Ivan Williams
  - Karen Siu
  - Marcus Kitchen
  - Konstantin Pavlov
  - Jeff Crosbie
  - Rob Lewis
  - Chris Hall
  - Sally Irvine
  - Michael Morgan
- Monash Physiology
  - Stuart Hooper
  - Megan Wallace
  - Melissa Siew
- Adelaide University
  - David Parsons
- Spring-8, Japan
  - K Uesugi
  - Naoto Yagi
  - M Suzuki

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Why do Imaging on Synchrotrons?

- The only way to get
  - Deep penetration
  - Micron spatial resolution
  - High sensitivity (soft tissue contrast), elemental specificity
  - Video time resolution
- Methods developed on synchrotrons can be translated to humans
  - We require some beamlines suitable for human studies

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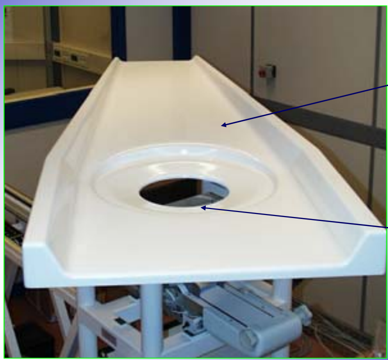
SYRMA project: patient support



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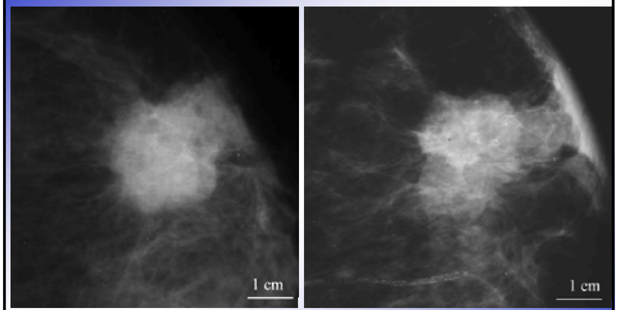
Fulvia Arfelli, Univ. Trieste

SYRMA: Clinical Images



Trieste Hospital (GE Senographe 2000D)

Synchrotron at 18 keV

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Fulvia Arfelli, University Trieste

Phase Contrast at Monash Medical Centre

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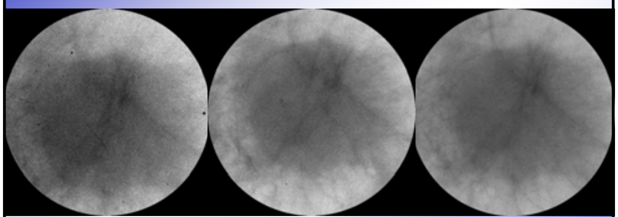
Ivan Williams

Detail



Contact Optimal 'Konica like'

Note: This is not a very good way to show a digital image.

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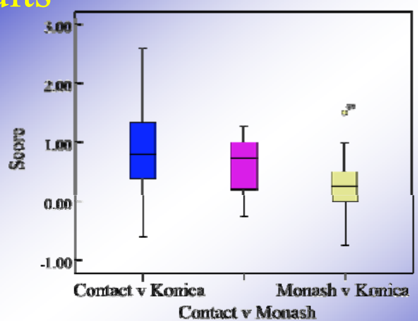
Results



Box-and-whiskers plot of the raw data averaged for each scorer for each of the three scoring comparisons, a positive score indicates that the second of the two geometries involved in the comparison was scored to advantage. The horizontal line within each box denotes median, box covers 25th percentile, whiskers denote the greater of 3.5 times 25th percentile and outer most point.
The two left-most show that two PCI geometries scored better than the Contact. The bar on the right shows that the Konica geometry was scored better than the Optimised. The single data point at 1.4 in the Optimised vs Konica comparison is an extreme outlier more than three standard deviations from the median.

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Ivan Williams

Future Prospects

- Drive is to understand dynamic processes
- Alternative imaging modalities will continue to improve
  - MRI, PET, SPECT, optical, IR
  - Can be placed in lab
- Synchrotron must concentrate on strengths
  - Combined high spatial and temporal resolution in 'thick' objects
  - We must move towards multi-modality imaging

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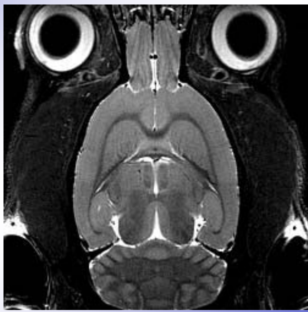
11.7T MRI

BioSpec 117/16 USR



Turbo Spin Echo (RARE) with 68 x 68 µm2 in plane resolution, 0.5 mm slice thickness.

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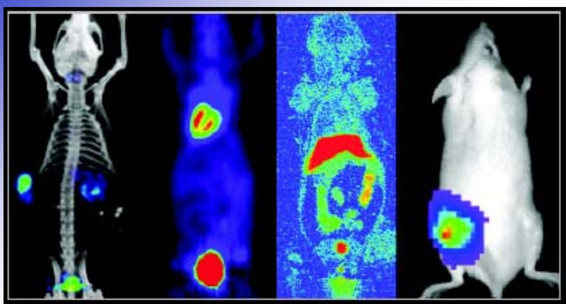
Methods



microPET-CT bioluminescence **microPET** **Autoradiography Optical**

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Robin Absorption and Phase

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