Coherent X-Ray Imaging and applications in computed tomography

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Summary

- Interest of phase contrast imaging
- Phase imaging techniques
- Spatial coherence
- Fresnel diffraction $\Leftrightarrow$ direct problem

- Edge enhancement
- Phase retrieval $\Leftrightarrow$ inverse problem
- Coherent Diffraction Imaging
- Nano-Probe Imaging
  - Combination Projection Microscopy – X-ray Fluorescence
  - 'Zoom' or Local Tomography
Transmission Imaging

Simple transmission

Absorption

Phase

Sample

• Dream 1: **Zero Dose**
  
  Increase the energy
  
  Absorption contrast ↓ replaced by phase contrast

• Dream 2: Improve the **Sensitivity**
  
  Absorption contrast too low
  
  high spatial resolution
  
  light materials
  
  similar attenuation
- Weak interaction with matter
- Refractive index $n$ (X-rays):

\[ n = 1 - \delta + i \beta \]

$\delta >> \beta$

$10^{-6}$ $10^{-9}$

\[ \beta = (\lambda /4\pi) . \mu \]

linear attenuation coefficient $\mu$

\[ \beta = \frac{r_c \lambda^2}{2 \pi V} \sum f'_p \]

$\delta$ – Refractive index decrement

- proportional to electron density
- inversely proportional to energy

\[ \delta = \frac{r_c \lambda^2}{2 \pi V} \sum (Z_p + f'_p) \]

$\approx 1.3 \times 10^{-6} \rho \lambda^2$

$\rho$ in g/cm$^3$, $\lambda$ in Å
Phase contrast vs Absorption

Gain of up to a few 1000!

Polymer sphere with two layers

- Inner layer: polystyrene thickness 30 µm
- Outer layer: parylene thickness 15 µm

Propagation

- Diameter: D = 83 cm

Absorption

\[ D = 0.03 \text{ m} \]

\[ \lambda = 0.7 \text{ Å} \]

200 µm
Transmission through a sample

refractive index \( n = 1 - \delta + i\beta \)

\[ \Delta z \quad e^{ik_{\text{medium}}\Delta z} = e^{i\frac{2\pi}{\lambda} n\Delta z} = e^{i\frac{2\pi}{\lambda} \Delta z} - i\frac{2\pi}{\lambda} \delta \Delta z - \frac{2\pi}{\lambda} \beta \Delta z \]

Transmission function
projection of the refractive index distribution

\[ u_{\text{inc}}(x,y) \quad u_0(x,y) = T(x,y). u_{\text{inc}}(x,y) \]

Amplitude
\[ A(x,y) = e^{-B(x,y)} = e^{-\frac{2\pi}{\lambda} \int \beta(x,y,z) \, dz} \]

Phase
\[ \varphi(x,y) = -\frac{2\pi}{\lambda} \int \delta(x,y,z) \, dz \]
Transmission Imaging \rightarrow Tomography

Parallel beam case ...

the whole object is imaged at once, slices are independent

Exit wave gives directly *projections* of the object

\[
B(x, y) = \frac{2\pi}{\lambda} \int \beta(x, y, z) \, dz \\
\varphi(x, y) = -\frac{2\pi}{\lambda} \int \delta(x, y, z) \, dz
\]
Phase Sensitive Techniques

Phase Retardation

\[ \Delta \varphi = -\frac{2\pi}{\lambda} \delta \cdot \Delta s \]

Deflection

\[ \Delta \alpha \sim \mu \text{rad} \]

Phase gradients

\[ \Delta \alpha = -\frac{\lambda}{2\pi} \frac{\partial \varphi}{\partial x} \]

At zero distance:

Intensity

\[ I_0 = |u_0|^2 = I_{\text{inc}} \cdot \exp(-\int \mu \, dz) \]

⇒ all phase information is lost
1) Interferometry

- Silicon crystal slices as beam splitter, deviator, recombiner
- Sensitive to the phase
- Too sensitive?
- Synchrotron: because of flux

Interference with reference beam
Ando & Hosoya, 1972
Bonse-Hart 1965; Momose
2) Differential Phase Contrast Imaging

**Analyser based imaging**

- Silicon crystals as collimator/monochromator and analyser
- Sensitive to the angular deviations
- Synchrotron: because of flux (and spatial coherence)

Zaumseil, 1980
Belyaevskaja & Ingal, Chapman

\[ \frac{\partial \Phi}{\partial x} \]
2) Differential Phase Contrast Imaging

Analyser based imaging

\[ \frac{\partial \phi}{\partial x} \]

Analyser as angular filter

Zaumseil, 1980

Belyaevskaya & Ingab, Chapman

Absorption
Radiograph of a white mouse

Belyaevskaya & Ingab
Physica Medica (1996) 12, 75
Ag K - radiation

Phase Dispersion
Radiograph
2) Differential Phase Contrast Imaging

Grating Based Phase Contrast Imaging

- phase grating as beam splitter
- absorption grating as transmission mask
- x-ray wavelength $\lambda \sim 0.1$ nm, grating periods $\sim 2-4$ $\mu$m
- demonstrated with laboratory source

$T. \text{ Weitkamp, F. Pfeiffer, Ch. David et al}$
2) Differential Phase Contrast Imaging

Grating Based Phase Contrast Imaging

- Phase object
- beam-splitter
- phase grating
- amplitude grating
- analyzer

\[ \frac{\partial \varphi}{\partial x} \]

- phase grating as beam splitter
- absorption grating as transmission mask
- x-ray wavelength \( \lambda \sim 0.1 \text{ nm} \), grating periods \( \sim 2-4 \mu \text{m} \)
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T. Weitkamp, F. Pfeiffer, Ch. David et al
2) Differential Phase Contrast Imaging

Grating Based Phase Contrast Imaging

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*T. Weitkamp, F. Pfeiffer, Ch. David et al*
2) Differential Phase Contrast Imaging

Human Brain Imaging

Energy: 23 keV
Pixel size: 5 μm
G1-to-G2 distance: 480 mm


*Limited spatial resolution
Very high density resolution*
3) Propagation

- Crystal as monochromator
- Synchrotron: because of spatial coherence + flux
- White beam variant: laboratory micro-focus X-ray sources neutrons

\[ \frac{\partial^2 \Phi}{\partial^2 x} + \frac{\partial^2 \Phi}{\partial^2 y} \]

Fresnel diffraction
Hartman (1994)
Cloetens; Snigirev; Wilkins

\[ \text{sample} \rightarrow D \rightarrow \text{detector} \]
3) Propagation

Fresnel diffraction

\[
\frac{\partial^2 \Phi}{\partial^2 x} + \frac{\partial^2 \Phi}{\partial^2 y}
\]

Hartman (1994)

Cloetens; Snigirev; Wilkins

Radiographs of a human vertebra

Absorption

Propagation

250 µm
Phase Contrast Imaging methods at ESRF

Analyzer-based PCI

Grating-based PCI

Propagation-based PCI

ID17
Medical Imaging
A Bravin

BM05/ID19
wavefront sensor
Ch David, F Pfeiffer
I Zanette / A Rack

ID19 / ID22NI
Edge enhancement
Holotomography
Coherence of a beam

**Temporal Coherence**
Correlation in time $u(t) u(t+\Delta t)$

- Monochromaticity: $\Delta \lambda / \lambda$
- Longitudinal coherence length: $\lambda^2 / \Delta \lambda$

**Spatial Coherence**
Correlation in space $u(x) u(x+\Delta x)$

- Angular source size: $\alpha$
- Transverse coherence length: $\lambda / 2\alpha$
Spatial Coherence

• Interference

monochromatic light

condition: \( l \) smaller than transverse coherence length

visibility is given by the complex degree of coherence \( \mu_{12}(x_1,x_2) \)

• Applications with X-rays
  • Phase Contrast Imaging / Coherent Diffraction Imaging
  • Speckle experiments
  • Extreme Focusing
Spatial Coherence

- Hard X-ray / neutron sources are \( \sim \) incoherent

- Wave is partially coherent when the source is small and far

\[
\alpha = \frac{z_2}{z_1} \]

- Transverse coherence length

\[
l_{coh} = \frac{\lambda}{2\alpha} = \frac{\lambda z_1}{2s}
\]

Feature size \( a (1/f) \)

\[
D_{opt} = \frac{a^2}{2\lambda}
\]

\[
|\mu_{12}(\lambda D f)|
\]

\[
l_{coh} > \frac{a}{2}
\]
Everything in the beam can act as a phase object

- Stringent requirements on beamline optics: Polished windows, Monochromator surface, ... Dust!

no thickness variations (surface) no density fluctuations (volume)
Diffraction at apertures

- Fresnel diffraction

![Diagram showing aperture and geometrical shadow with distances $z_1$ and $z_2$.]

- $\lambda = 0.7$ Å
- $100$ µm
- $z_1 = 140$ m
- $z_2 = 6$ m

$\Rightarrow$ aperture close to sample
Experimental Setup: Parallel Beam

Implemented on ID19 / ESRF

Dedicated microtomograph

Monochromator:
- double Si crystal ($\Delta\lambda/\lambda=10^{-4}$)
- or multilayer ($\Delta\lambda/\lambda=10^{-2}$)

Sample stage
rotation stage (tomography)

Detector
- CCD-based
- on translation (propagation)
  crucial for spatial resolution
Simple Propagation

- Fresnel diffraction over a distance $D$ (few mm to several meters)
  Defocusing in electron microscopy
  In-line Gabor holography

- At least
  \[
  \frac{\partial^2 \Phi}{\partial^2 x} + \frac{\partial^2 \Phi}{\partial^2 y} \neq 0
  \]
  no contrast
  contrast
Fresnel diffraction

Fresnel Integral (1D)

\[ u_D(x) = \frac{1}{\sqrt{i \lambda D}} \int_{-\infty}^{+\infty} u_0(x_0) \exp\left[-i \frac{\pi}{\lambda D} (x - x_0)^2\right] dx_0 \]

- Convolution in real space
- Multiplication in reciprocal space
- Small angles \hspace{1cm} \text{paraxial approximation}
**Fresnel diffraction**

\[ D \]

Example at \( \lambda = 0.5 \text{Å} \) (25 keV):

- \( a = 1 \ \mu\text{m} \) \( \Rightarrow \) \( D = 10 \text{ mm} \)
- \( a = 40 \ \mu\text{m} \) \( \Rightarrow \) \( D = 16 \text{ m} \)

- *In principle*: complete object contributes to a point of the image
- *In practice*: only finite region: first Fresnel zone

**radius**  
\[ r_F = \sqrt{\lambda D} \]

- First Fresnel zone determines the *sensed lengthscale*
- Distance to be most sensitive to object with size \( a \)  
  \[ D_{opt} = \frac{a^2}{2\lambda} \]
Fresnel diffraction in reciprocal space

Propagation over distance $D$ along $z$: Simple phase factor given by: $\exp(i \, k_z \, D)$

each plane wave component is modified in its phase: multiplication with propagator (function of spatial frequency $f$): 

$$P_D(f) = \exp[-i \, \pi \, \lambda \, D \, |f|^2]$$
Contrast Transfer Functions

- Fourier Transforms of the intensity and phase are linearly related

\[ I_D(f) = \delta_{\text{Dirac}}(f) - 2 \cos(\pi \lambda D f^2) \cdot B(f) + 2 \sin(\pi \lambda D f^2) \cdot \varphi(f) \]

amplitude contrast factor phase contrast factor
valid in case of a slowly varying phase, weak absorption
Contrast transfer functions

Decreasing linewidth
Increasing spatial frequency

Object invisible!
Contrast depends strongly on period or spatial frequency

period ≈ 720 nm
period ≈ 530 nm
period ≈ 610 nm
Fresnel diffraction: specificities

Compared to absorption imaging \((D = 0)\):

- no simple relationship between object and image (Fresnel pattern)
- finite region contributes to each point (convolution effect)
  \[\Rightarrow\text{inverse problem not straightforward}\]

Compared to Fraunhofer diffraction (far field, \(D \to \infty\)):

- **far field:** diffraction pattern only depends on diffraction angles, not on distance
  \[\sqrt{\lambda D} \gg a\]
  - intensity proportional to squared modulus of Fourier transform of the object
  \[\Rightarrow \text{PHASE PROBLEM} \text{ (needs other information)}\]

- **Fresnel diffraction:** diffraction at finite distance
  \[\sqrt{\lambda D} \approx a\]
  - pattern strongly depends on distance
  \[\Rightarrow \text{helps to solve the phase problem}\]
Simulation

Direct Space

- Multiplication wave with transmission function

Reciprocal Space

- Convolution

OBJECT

\[ T = e^{-B} e^{i\varphi} \]

PROPAGATION

- Diffraction integral

INTENSITY

- Multiplication FT wave with propagator

\[ P_D = \exp(-i\pi\lambda Df^2) \]

COHERENCE / DETECTOR

- Convolution intensity with projected source PSF detector

- Multiplication FT intensity with degree of coherence detector transfer function

\[ |\mu_{12}(\lambda Df)| \]

\[ R(f) \]
Porosity in Quasi-crystals

Hole = dodecahedron seen along the 2-fold axis

λ = 0.52 Å
D = 0.4 m

Simulations

Phase imaging of lipid membranes in solution

A Beerlink, T Salditt et al., Soft Matter, 2012, 8, 4595
Phase imaging of lipid membranes in solution

Native bilayer!
Thickness of about 3 nm

Fresnel fringes provide access to local bilayer structure

A Beerlink, T Salditt et al., Soft Matter, 2012, 8, 4595
Phase imaging of lipid membranes in solution

Possibility to study dynamic interface processes in solution
Electrically excited black lipid membrane in micro-fluidic device

AC field
(1 Hz freq., 70 mV amp.)

A Beerlink, T Salditt et al., Soft Matter, 2012, 8, 4595
edge detection versus holography (Fresnel diffraction)

each edge imaged independently
no access to phase, only to border
\[ \sqrt{\lambda D} \ll a \]

access to phase, if recorded at \( \neq D \)'s
defocused image
\[ \lambda = 0.7 \text{ Å} \]
\[ 50 \text{ µm} \]
\[ \sqrt{\lambda D} \approx a \]
**Edge Enhancement**

- Essentially edge enhancement
  Weak defocusing (and weak contrast!)

\[ \sqrt{\lambda D} \ll a \]

- Radiography (2D)
  \[ I_D(x, y) \approx I_0(x, y) \cdot \left[ 1 - \frac{\lambda D}{2\pi} \Delta_{xy} \varphi(x, y) \right] \]

- Tomography (3D)
  \[ o(x, y, z) \approx \mu(x, y, z) - D \Delta_{xyz} \delta(x, y, z) \]

- Detection of cracks, holes, reinforcing fibers, particles

The European Light Source
Edge Enhancement

Absorption Contrast: Small sample-detector distance

Phase Contrast: Large sample-detector distance
Virtual slices through heads of tiny staphylinid beetles

**Phase Retrieval**

**Flemish School:** D. Van Dyck, P. Cloetens, JP Guigay  
Transfer Function Approach / Focus variation

- Series of images recorded at different distances
- Each distance is most sensitive to a specific range of spatial frequencies

**Australian School:** K. Nugent, T. Gureyev, D. Paganin  
TIE (transport of intensity)

\[
\frac{\partial I}{\partial z} = -\frac{\lambda}{2\pi} \nabla (I \nabla \varphi)
\]
Paganin Approach (TIE variant)

- Single distance reconstruction for **homogeneous objects**
  Valid for **short distance, arbitrary absorption (!) and constant δ/β**

- Pro's:
  - Single distance
  - Most correct of simple approaches (arbitrary absorption)
  - Robust: homogeneity assumption has regularizing effect

- Contra's:
  - Weak distance, weak contrast → contrast is not optimized
  - Not adapted to very inhomogeneous objects
  - Low-pass filter → resolution is deteriorated

\[
\int \mu \, dz = -\ln \left( FT^{-1} \left\{ \frac{FT(I_D)}{\pi \lambda D f^2 \frac{\delta}{\beta} + 1} \right\} \right)
\]
Paganin Approach (TIE variant)

Example of implementation: ANKA phase, ImageJ plugin

Phase Retrieval: multiple distances

- non-absorbing foam
- 4 images recorded
- $E = 18$ keV

\[ \lambda = 0.21 \text{ m} \]
\[ D = 0.51 \text{ m} \]
\[ D = 0.90 \text{ m} \]

Least squares minimization of cost functional

\[ K = \sum_{m} |I_{m}^{\exp}(f) - I_{m}^{calc}[\varphi(f)]|^2 \]

50 $\mu$m
Holo-tomography

1) phase retrieval with images at different distances

2) tomography: repeated for $\approx 1000$ angular positions

3D distribution of $\delta$ or the electron-density improved resolution straightforward interpretation processing
Phase retrieval: transfer function approach

\[ I_m(f) = 2 \sin(\pi \lambda D_m f^2) \varphi(f) \]

\[ \varphi(f) = \frac{\sum_m \sin(\pi \lambda D_m f^2) I_m(f)}{\sum_m 2 \sin^2(\pi \lambda D_m f^2)} \]

Non-iterative (fast!)

\[ \sin(\pi \lambda D f^2) \]

⇒ Optimization of the choice of distances

\[ \frac{1}{N} \sum_m \sin^2(\pi \lambda D_m f^2) \]

4 distances
Phase Tomography of Arabidopsis seeds

3D structure of Arabidopsis seeds in their native state
- wet sample, no preparation
- no staining, no fixation, no cutting, no cryo-cooling

Holotomographic approach
Contrast proportional to the electron density
4 distances, 800 angles
$E = 21 \text{ keV}$

Phase Tomography of Arabidopsis seeds

In situ 3D imaging of a seed of an Arabidopsis plant

wet sample, no preparation

Radiograph  $D = 10 \text{ mm}$  Spectrum – Fourier transform

Contrast factor

50 $\mu\text{m}$
Phase Tomography of Arabidopsis seeds

In situ 3D imaging of a seed of an Arabidopsis plant

wet sample, no preparation

Radiograph  $D = 30 \text{ mm}$  Spectrum

Contrast factor

50 µm
Phase Tomography of Arabidopsis seeds

In situ 3D imaging of a seed of an Arabidopsis plant

wet sample, no preparation

Radiograph \( D = 60 \text{ mm} \)

Spectrum

Contrast factor

50 \(\mu\text{m}\)
Phase Tomography of Arabidopsis seeds

In situ 3D imaging of a seed of an Arabidopsis plant

wet sample, no preparation

Radiograph \( D = 100 \text{ mm} \)

Spectrum

Contrast factor

50 \( \mu \text{m} \)
Phase Tomography of Arabidopsis seeds

Holotomographic approach
Four distances
E = 21 keV

Seed of Arabidopsis

Tomographic Slices

30 µm Cotyledon
P Cloetens, R Mache, M Schlenker, S Lerbs-Mache, PNAS (2006) 103, 14626
Phase Tomography of Arabidopsis seeds

Seed of Arabidopsis

Tomographic Slice

10 µm

10 µm

5 µm

10 µm

5 µm

protoderm

tegumen

intercellular spaces
organites
(protein stocks)
Phase Tomography of Arabidopsis seeds

three-dimensional network of intercellular air space

gas exchange during germination
and/or
rapid water uptake during imbibition

Role?

Skull and brain of a 300 million-year-old chimaeroid fish revealed by synchrotron holotomography.

Life Sciences: Evolution

A Pradel et al, PNAS, 106, 5225 (2009)
Life Sciences: Evolution

A Pradel et al, PNAS, 106, 5225 (2009)
Discovery of oldest known preserved brain
Coherent Diffraction Imaging

Far field diffraction pattern ($D \to '\infty'$)
Lensless Imaging / Diffraction microscopy / CDI
Non-periodic object
Interest: high spatial resolution

\[ \sqrt{\lambda D} \gg a \]
Iterative phase retrieval algorithm

R.W. Gerchberg & W.O. Saxton, Optic (1972) 35, 237

http://xray1.physics.sunysb.edu/research/intro6.php#contents
Coherent Diffraction Imaging (CDI)

Three-dimensional visualization of a human chromosome
Nishino et al, PRL 102, 018101 (2009)
Unstained, chemically fixed, air-dried sample

RIKEN, Spring-8
Coherent scanning X-ray diffraction alias Ptychography: applicable to extended objects using diffraction patterns of overlapping regions.

Amplitude and phase of the reconstructed region.
Ptychographic X-ray computed tomography

Experiments at cSAXS beamline SLS
M. Dierolf, … F. Pfeiffer,

Osteocyte lacunae and the interconnective canalicular network in bone
**Scanning Transmission Microscopy - STXM**

X-ray Fluorescence: element distribution Slow; trace elements XAS, XEOL, Diffraction

**Projection Microscopy - PXM**

electron density Dose efficient, fast; magnified holo(tomo)graphy
**The Nano-Imaging end-station ID22NI**

- **Focus size**: 43 nm (v) x 59 nm (h) (17 keV)
- **Flux**: $\sim 10^{12}$ ph/s
  - $\rightarrow 4 \times 10^8$ ph/s/nm$^2$ or $4 \times 10^6$ ph/s/Å$^2$
- **Energy range**: 17-29 keV
- **Tunable bandwidth**: $\Delta E/E \sim 1.5 - 7\%$

**Multilayer KB Optics**
Projection Microscopy (PXM)

Plane wave illumination

Object

Detector

Limit cases

\[ z_1 \gg z_2 \]

\[ D = z_2 \; ; \; M = 1 \]

\[ z_1 \ll z_2 \]

\[ D = z_1 \; ; \; M = z_2/z_1 \]

Prop. by D Van Dyck, J Spence

\[ D = \frac{z_1 \cdot z_2}{z_1 + z_2} \]

Magnification
Magnified in-line holograms of Au Xradia test pattern

$E = 17.3$ keV

$D \approx 44$ mm

Projection Microscopy
Magnified in-line holograms of Au Xradia test pattern
$E = 17.3$ keV

$D \approx 34$ mm
Projection Microscopy

Magnified in-line holograms of Au Xradia test pattern
$E = 17.3$ keV
Projection Microscopy

Magnified in-line holograms of Au Xradia test pattern
$E = 17.3$ keV

$D \approx 29$ mm
Projection Microscopy: phase retrieval

1) Linear, direct method
   Linear least squares solution

\[ \tilde{\Phi}(f) = \frac{\sum_m K_m(f) \cdot \tilde{I}_m(f)}{\sum_m K_m^2(f) + \epsilon} \]

+ 2) Iterative improvement (optional)
Projection Microscopy: phase retrieval

X-radia gold test pattern
Innermost line width: 50 nm
Energy = 17.3 keV
Field of view: 80 µm
Pixel size: 53 nm

Au Fluorescence; 25 nm
Nano-particles for cancer therapy

Poly-functional nano-particles for cancer therapy

Quest for *Magic Bullet Nanoparticles*:
- Potential for **in vivo imaging** of targeted tumour cells
- Increased **target selectivity for cancer cells**
- Increase the dose **efficacy of therapeutic radiation**
- Ability to **inhibit DNA damage responses** in targeted cells

- **Limited (geno-)-toxicity**

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Dr. B. Kysela  
Genome Stability and DNA Repair Group  
The Medical School, University of Birmingham  
Project: MRC grant support

D Lewis et al, Nanomedicine, 5, 1547-1557 (2010)
X-ray fluorescence nano-imaging of internalized Lanthanide nanovectors

Pt nano-particles target the nucleus

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Project: MRC grant support

D Lewis et al, Nanomedicine, 5, 1547-1557 (2010)
Combined Phase Contrast and XFM

Phase $\rightarrow$ Projection of the electron / mass density
+ X-ray Fluorescence $\rightarrow$ Areal mass of specific elements
↓

Mass fractions (averaged over the thickness)
Better Mass normalization
Combined Phase Contrast and XFM

Combined Phase Contrast and XFM

Bias due to heterogeneous sample thickness is removed

Phase contrast in a scanning X-ray microprobe

Alternative approach: Differential phase contrast with a segmented detector in a scanning X-ray microprobe

Transmission detector measures:
- Absorption
- Deflection of beam due to specimen

Hornberger et al, JSR 15, 355 (2008)
Zoom Tomography

- Zoom 'non-destructively' into a region of interest of a tissue, cell, ...
- Keep sample preparation and measurement straightforward
- Sample size >> field of view

R Mokso, P Cloetens, E Maire, W Ludwig, JY Buffière, APL, 2007, 90, 144104
Phase NanoCT of human bone ultrastructure

Interstitial tissue

Osteonal tissue

M Langer, A Pacureanu, H Suhonen, P Cloetens, F Peyrin, PLoS ONE
Phase NanoCT of human bone ultrastructure

Energy = 17 keV
Phase Retrieval using 4 distances
2999 projections over 360 deg
60 nm voxel size, 90 µm field of view

M Langer, A Pacureanu, H Suhonen, P Cloetens, F Peyrin, PLoS ONE
Phase NanoCT of human bone ultrastructure

- Cement line
- Osteocyte Lacuna
- Canaliculi

• 3D lacuno-canalicular network imaged over several cells

Cm, On, Ca, Lc, It, Lt
Phase NanoCT of human bone ultrastructure

- 3D lacuno-canalicular network imaged over several cells
Phase NanoCT of human bone ultrastructure

- 3D lacuno-canalicicular network imaged over several cells
- 3D bone mineralization at the nanoscale
- The cement line is hyper-mineralized
Collagen fibers are organized as a twisted plywood structure in 3D.
Coherent X-rays allow to perform, in an easy way, phase (sensitive) imaging through propagation.

The coherence puts stringent requirements on all beamline optics.

Quantitative mapping of the phase (2D) and density (3D) is possible by combining images at different distances with an adapted numerical algorithm.

Coherent Diffraction Imaging has the potential to improve the spatial resolution beyond the limits of X-ray optics.

Nano-probe Imaging combines in a single setup Projection Microscopy [electron density / μ-structure] and Fluorescence Imaging [(trace) element distribution].
Some references