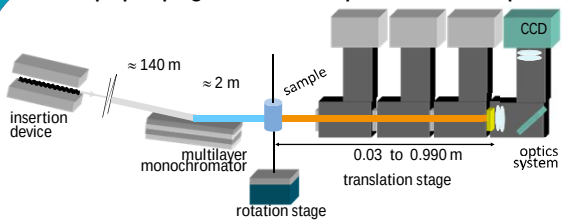
Comparison of regularized phase retrieval algorithms for X-ray phase contrast tomography – application on experimental data

L. Weber[1,2], M. Langer[1,2], P. Cloetens[2], F. Peyrin[1,2].

1- Créatis, Université de Lyon, CNRS UMR5220, Inserm U1044, INSA-Lyon, Université Lyon 1, Villeurbanne, France.
2- European Synchrotron Radiation Facility, Grenoble, France.

In-line phase contrast tomography

Setup: propagation-based parallel beam phase micro- CT



- X-ray phase contrast imaging offers several advantages versus absorption X-ray imaging
- With coherent X-rays, a simple experimental setup based on **propagation** can be used [1]
- Phase contrast is achieved by moving the detector downstream
- The object is modelled by the **complex refractive index** $n=1-\delta+i\beta$, where δ is the refractive index decrement and β the absorption index
- It depends on the material and the energy of the incoming beam. For soft tissue, at low energies (as used in radiology), δ is three order of magnitude larger than β
- Propagation modelled by **Fresnel diffraction**

- In holotomography [1], images are acquired at:
  - **several source-to-sample distances**
  - **under different angles of rotations**
- **Phase projections** are obtained by combining these multi-distance projections for each angle θ using phase retrieval.

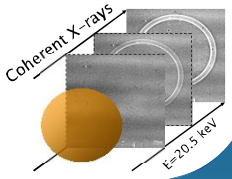


Image reconstruction

- **Modeling of the Direct Problem**
  - Attenuation B and phase shift ψ induced by the object described as projections perpendicular to the propagation direction z , with $\mathbf{x}=(x,y)$

    $$B(\mathbf{x})=\frac{2\pi}{\lambda}\int \beta\, dz \text{ and } \psi(\mathbf{x})=\frac{2\pi}{\lambda}\int \delta\, dz$$
  - At each angle θ, the interaction between the objet and the X-ray can be described as a transmittance function

    $$T_\theta(\mathbf{x})=\exp[-B_\theta(\mathbf{x})+i\,\psi_\theta(\mathbf{x})]$$
  - The free space propagation over a distance D can be modeled by the Fresnel transform involving the propagator P_D

    $$T_{\theta,D}(\mathbf{x})=(T_\theta * P_D)(\mathbf{x})$$
  - The intensity recorded by the detector at a distance D is

    $$I_{\theta,D}(\mathbf{x})=|T_{\theta,D}(\mathbf{x})|^2$$

- **Inverse problem**
  - Can be split in two steps
  - **Phase retrieval:** calculate the phase shift $\psi_\theta(\mathbf{x})$ from the phase contrast images at different propagation distances $\{I_{\theta,D}(\mathbf{x}), D=D1...Dn\}$

    ➔ Ill-posed inverse problem, which requires regularization.

    $$\hat{\psi}_\theta(\mathbf{x})=\arg\min_\psi \sum_D [\,|\,T_{\theta,D,\psi}(\mathbf{x})-I_{\theta,D}(\mathbf{x})|^2+\alpha|\psi_\theta(\mathbf{x})-\psi_{\theta,0}(\mathbf{x})|^2\,]$$
  - **Tomographic reconstruction:** reconstruct the 3D refractive index image from the phase shifts $\hat{\psi}_\theta(\mathbf{x})$ at different projection angles
  - Usually solved by filtered back-projection.

Regularized phase retrieval

Design of prior

- No prior (i.e. $\psi_{\theta,0}(\mathbf{x})=0$) induces low-frequency noise.
- Possible introduction of **prior** information to alleviate this problem.
- **Homogeneous** object: prior is designed assuming that phase is proportional to the attenuation (δ/β-ratio is constant) and applied in the projection domain [3].

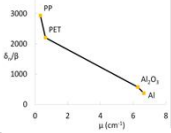
$$\psi_{\theta,0}(\mathbf{x}) = f(\mathbf{x}) * \frac{\delta}{2\beta}\ln[I_{\theta,0}(\mathbf{x})]$$

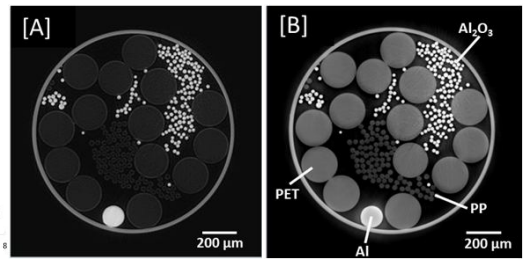
- For **heterogeneous** object of known material, the prior can be designed using several δ/β-ratios:
  - ➢ by thresholding the attenuation scan [4]
  - ➢ by assuming that the δ/β-ratio is a (continuous) function of the attenuation coefficient [5]

Application on phantom [5]

- Phase retrieval using homogeneous [A] and heterogeneous [B] prior.
- The sample is a phantom composed of multiple wires (Al, PET and PP and Al2O3), mounted inside a glass capillary.

- **Experimental conditions**
  - 1500 projections
  - 22.5 keV
  - 4 distances
  - Pixel size = 0.7 µm
  - Field of view = 1.4 mm



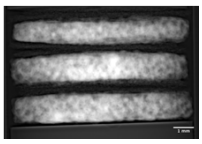


Application on 3D cell cultures

Introduction [6]

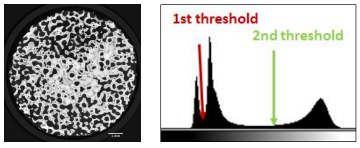
- Biomaterial (Skelite®) that can be used for 3D cell culture and artificial bone scaffolds in surgery
- Composed of Si-TCP(67%) and β-TCP/HA (33%)
- 3 scaffolds imaged: discs of 9 mm diameter, seeded with osteoblasts

- **Experimental conditions**
  - 2000 projections
  - 30 keV
  - 4 sample-to-detector distances
  - Pixel size=5 µm
  - Field of view = 10.2 mm

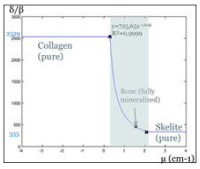


Regularization

- Reconstruction using the **homogeneous approach** (method 1) allows quantification of scaffolds but causes artifacts in soft tissue [7]
- We applied 3 **heterogeneous objects approaches** to improve soft tissue imaging
  - using thresholding (1st and 2nd threshold correspond to methods 2 and 3 respectively)
  - using a functional relationship between the attenuation coefficient and δ/β values (method 4)

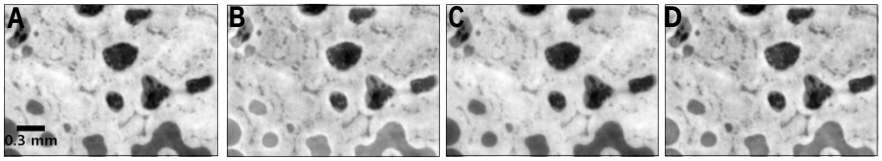


Example of a filtered slice of the attenuation scan, and the corresponding histogram; the histogram is clearly tri-modal, so that 2 different thresholds are possible. They were used in method 2 and method 3.

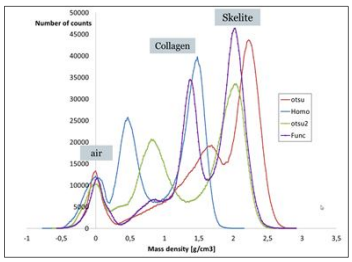


Functional relationship between the attenuation and the δ/β values, used in method 4

Results

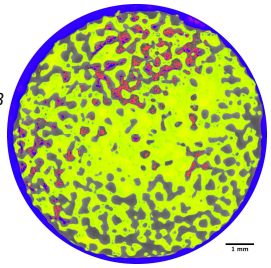


Crop of a reconstructed slice, using homogeneous approach [A], heterogeneous approach with 1st threshold [B] or 2nd threshold [C], and heterogeneous approach with a functional relationship [D].



Histograms of mass densities of the reconstructed slice with the 4 methods

Thresholded slice reconstructed with method 3



- **Theoretical mass densities:**
  - ρ(air)=1.3*10-3 g/cm3
  - ρ(Collagen)=1 g/cm3
  - ρ(Skelite®)=1.8 g/cm3

- Only 3 modes are visible using methods 1 and 2, while we can distinguish 4 modes with methods 3 and 4.
- Method 1 presents fringes around soft tissue
- Phase contrast artefacts are visible using methods 2 and 4.
- Method 3 seems more quantitative, and image quality is improved with regards to the other methods.

Conclusions

- Heterogeneous object approaches can clearly improve image quality
- Choice of good parameters (threshold, function) is of crucial importance

Perspectives

- Introduce spatial constraints in prior, i.-e. reconstruction while getting rid of the glass capillary
- Further quantitative studies using a calibrated phantom for example.

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Acknowledgements

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