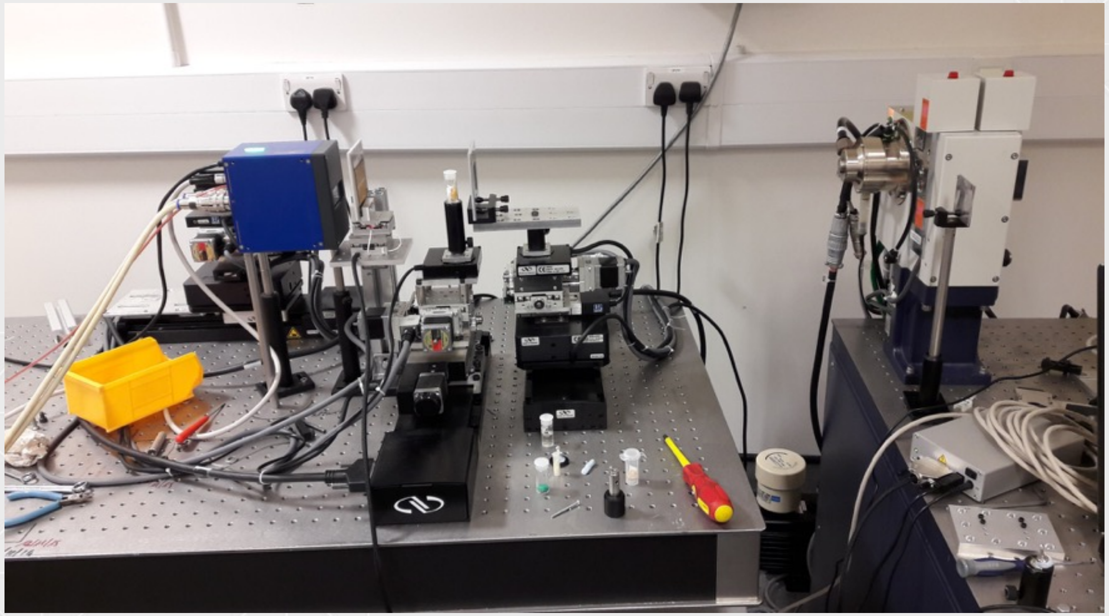
Laboratory-based x-ray phase contrast imaging for NDT and industrial testing



Sandro Olivo, Spokesperson, AXIm Group

(https://www.ucl.ac.uk/medical-physics-biomedical-engineering/research/groups-and-centres/advanced-x-ray-imaging-group-axim)

Medical Physics and Biomedical Engineering, UCL

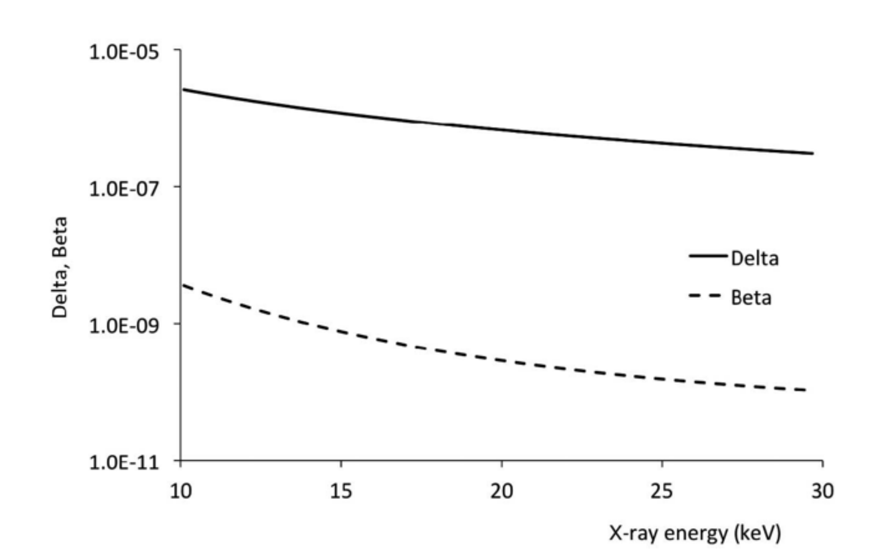
x-ray phase contrast imaging

$$n = 1 - \delta + i\beta,$$

$$\mu = \frac{4\pi\beta}{\lambda}$$

$$\delta = \frac{r_e \rho_e \lambda^2}{2\pi}$$

$$\Phi(x,y) = \frac{2\pi}{\lambda} \int_{\mathrm{object}} \delta(x,y,z)\mathrm{d}z,$$



Note 1) ~ 3 orders of magnitude larger 2) decreases more slowly with x-ray energy

x-ray phase contrast imaging

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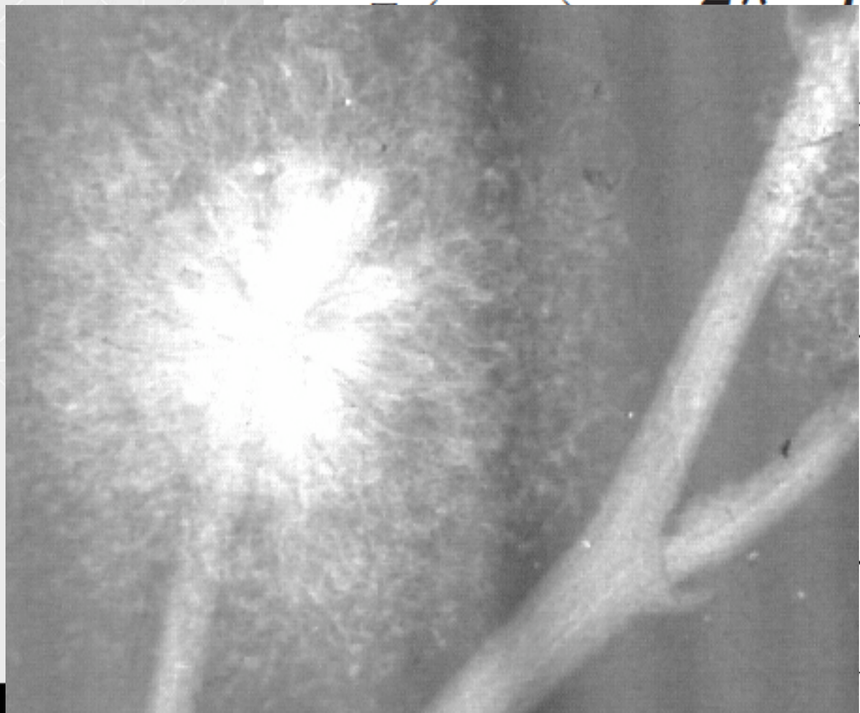
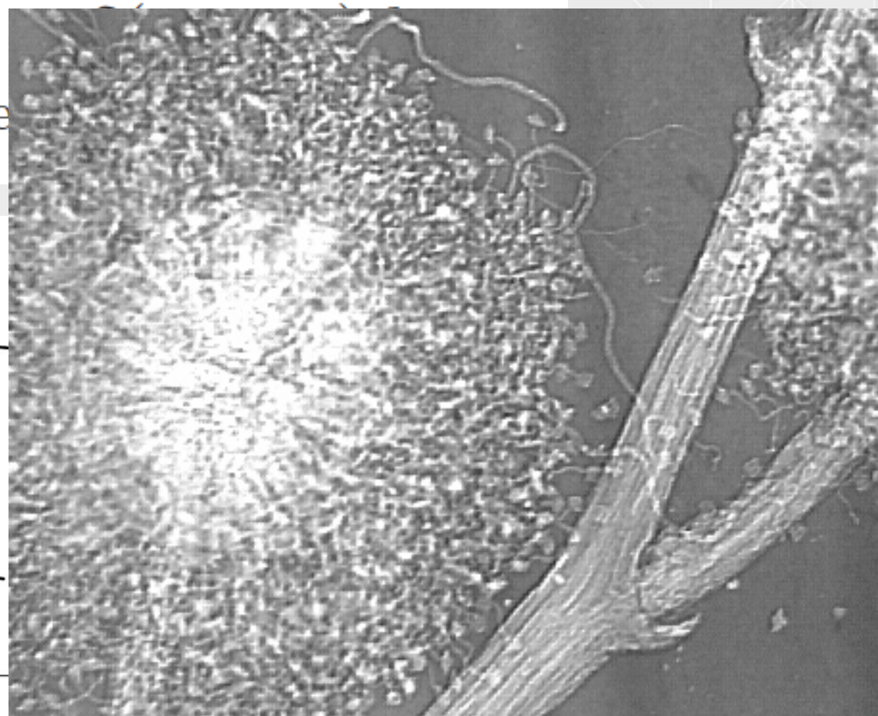




20

X-ray energy (keV)

Phase changes cause small angular deviations (x-ray refraction)

When crossing an object with negligible absorption (β~0)
but with $\delta \neq 0$, the X-ray wavefield changes from

$$\Psi = \Psi_0 \exp(-ikz)$$

to

$$\Psi = \Psi_0 \exp[i(-kz + \phi)]$$

with

$$\phi(x,y) = -r_e \lambda \int_{object} \rho_e(x,y,z)dz$$

(r_e classical electron radius, λ incident radiation wavelength, ρ_e electron density)

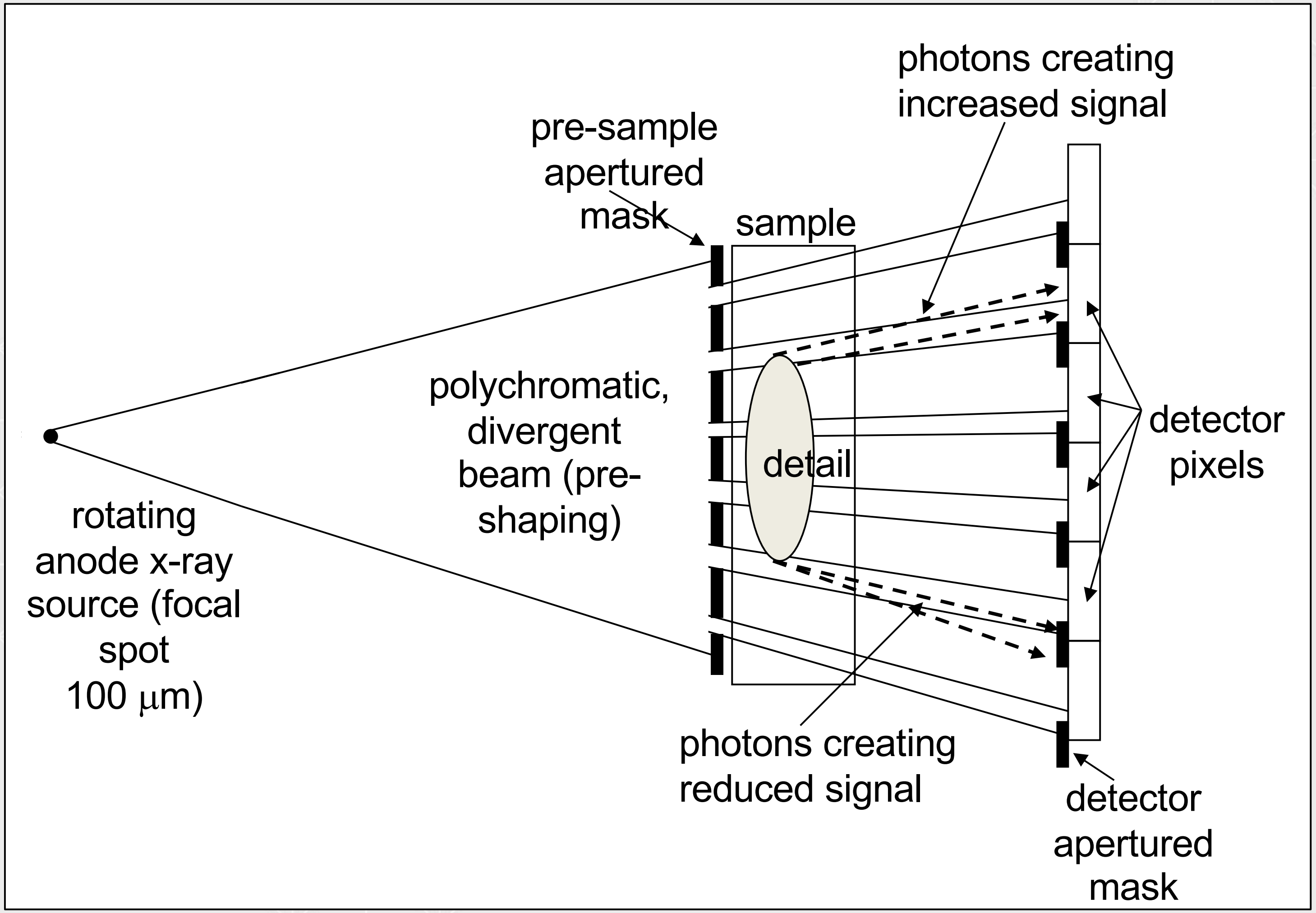
The new wavevector is therefore:

$$\vec{k}' = \left(\frac{\partial\phi}{\partial x}, \frac{\partial\phi}{\partial y}, \frac{2\pi}{\lambda}\right) = \vec{\nabla}_{xy}\phi + k\hat{z}$$

and the angular deviation (relative to the Initial propagation direction) is given by:

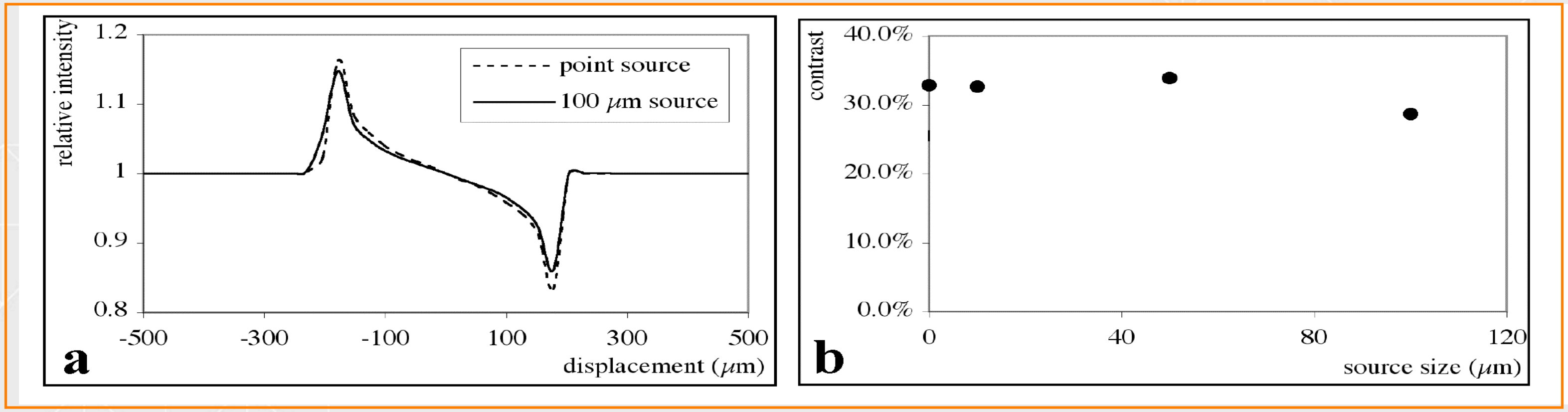
$$\alpha \cong \frac{1}{k}\left|\vec{\nabla}_{xy}\phi(x,y,\lambda)\right|$$

EDGE-ILLUMINATION XPCI with DIVERGENT and POLYCHROMATIC (=conventional) SOURCES



Olivo and Speller Appl. Phys. Lett. 91 (2007) 074106

Little loss of signal intensity for source sizes up to 100 μm



Which can be achieved with state-of-the-art mammo sources

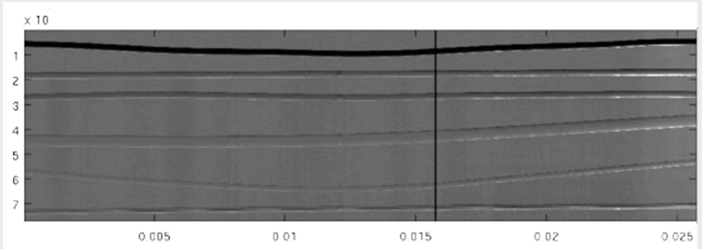
Why?

1) Because we are only relying on refraction, which survives under relaxed coherence conditions;
2) Because we are use aperture pitches matching the pixel size, i.e. BIG: the projected source size remains < pitch, and therefore blurring does "not" occur.

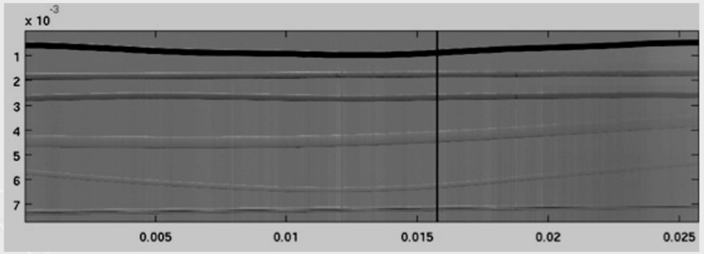
Olivo and Speller Phys. Med. Biol. 52 (2007) 6555-73

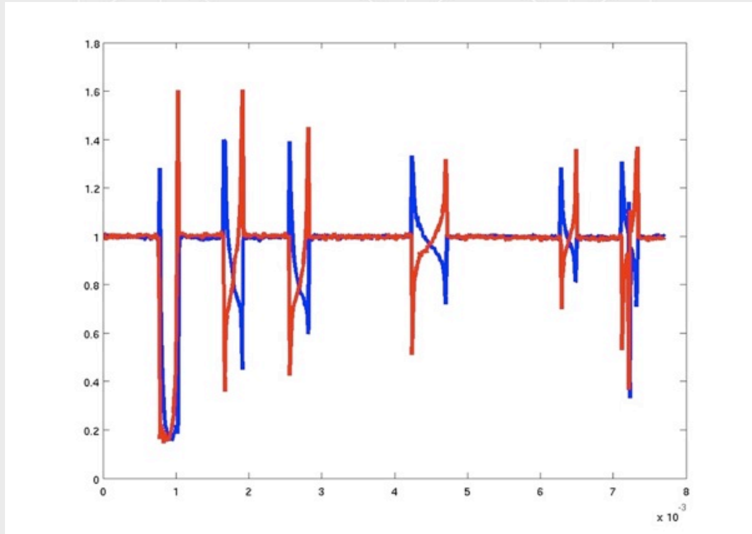
Quantitative phase contrast imaging

"SLOPE -"

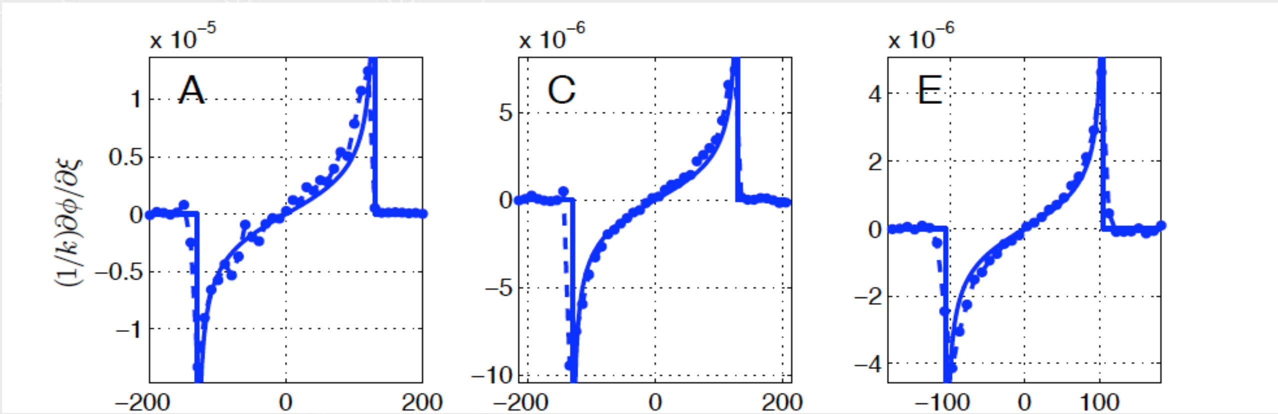


"SLOPE +"





Titanium Aluminum PEEK

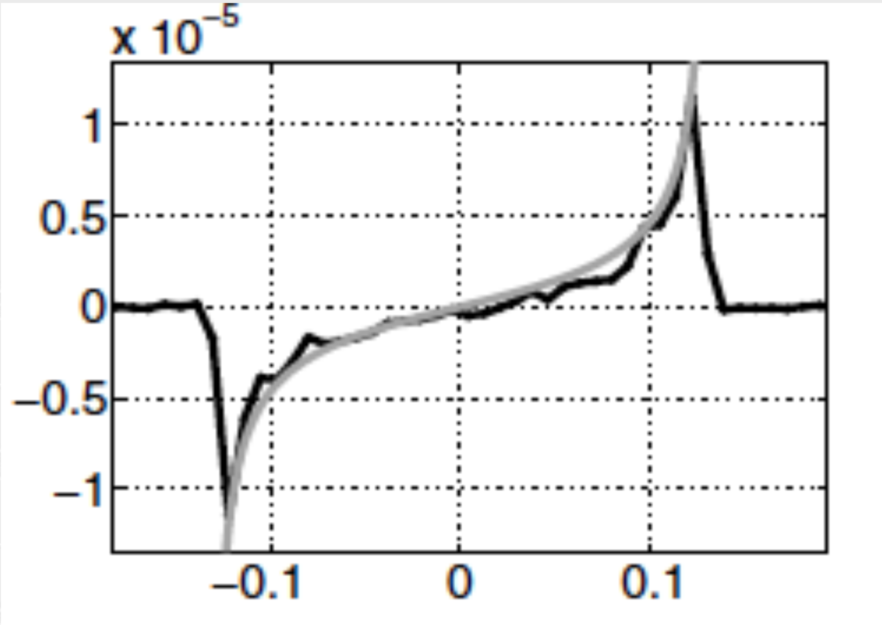


Highly precise retrieval, for both high and low Z materials, up to high gradients where other methods break down

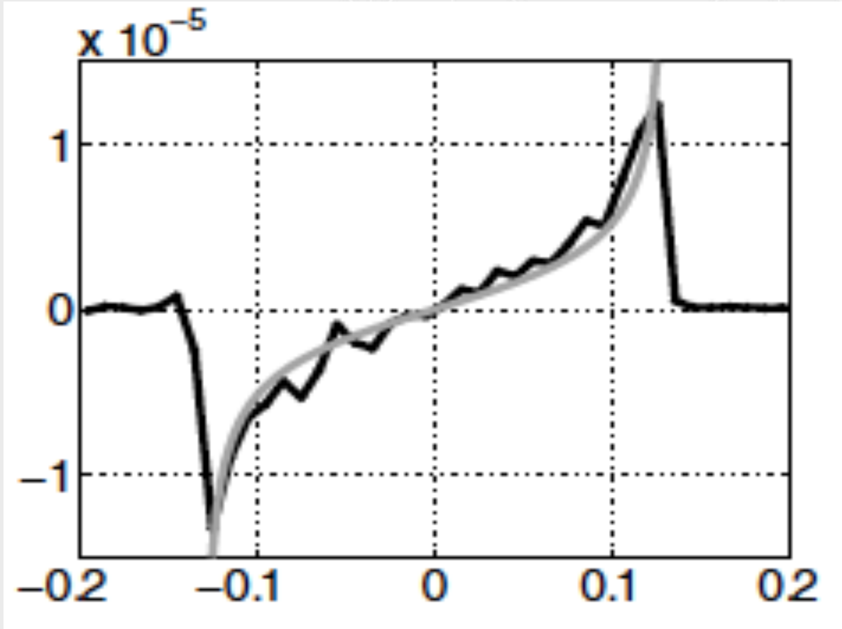
Munro *et al* Opt. Exp. 21 (2013) 647-61

Phase retrieval with synchrotron and conventional sources:

Ti filament: retrieved @ synchrotron



and with conventional source!



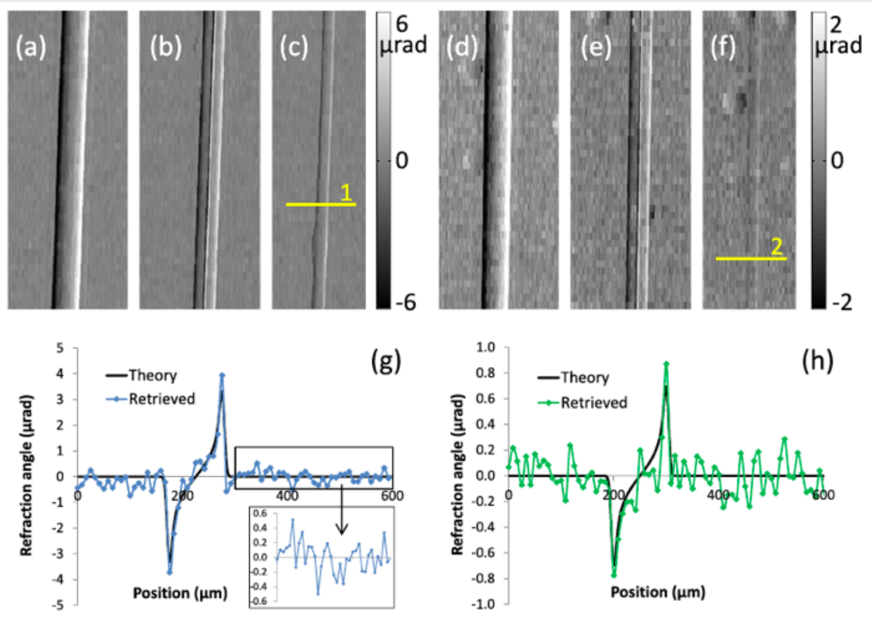
@ conventional source: incoherence modelled as beam spreading – the movement of the "spread" beam is then tracked and referred back to the phase shift that caused it.

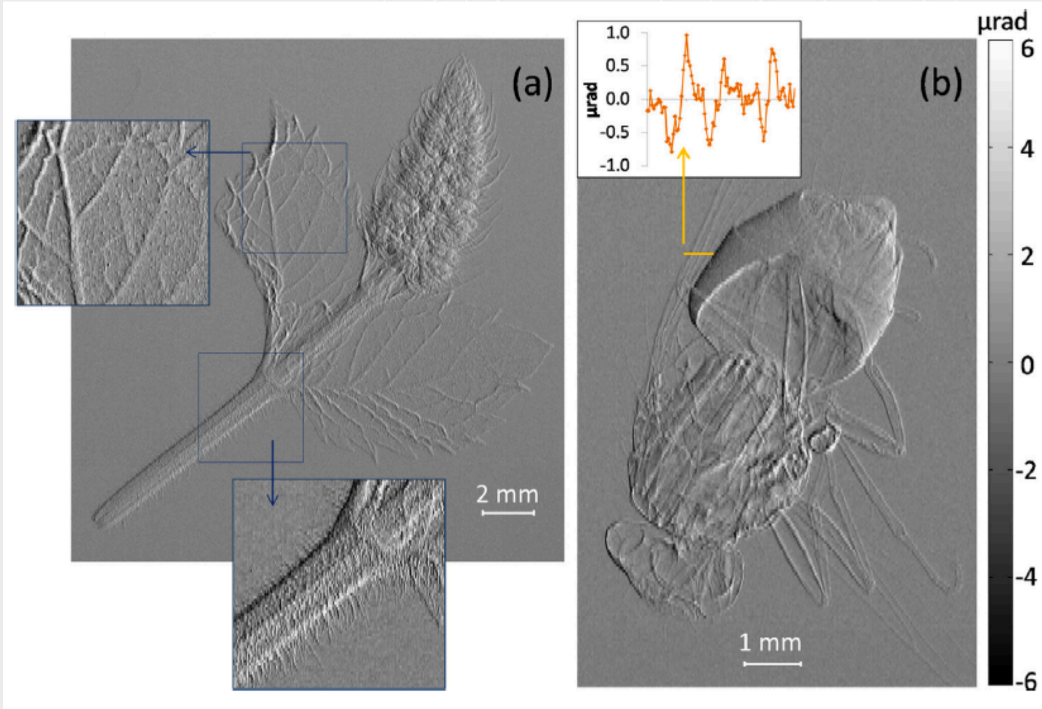
But with lots of care as far as "effective energy" is concerned!

(See Munro & Olivo Phys. Rev. A **87** (2013) 053838)

Munro *et al*, *PNAS* 109 (2012) 13922-7

sensitivity of the lab system:





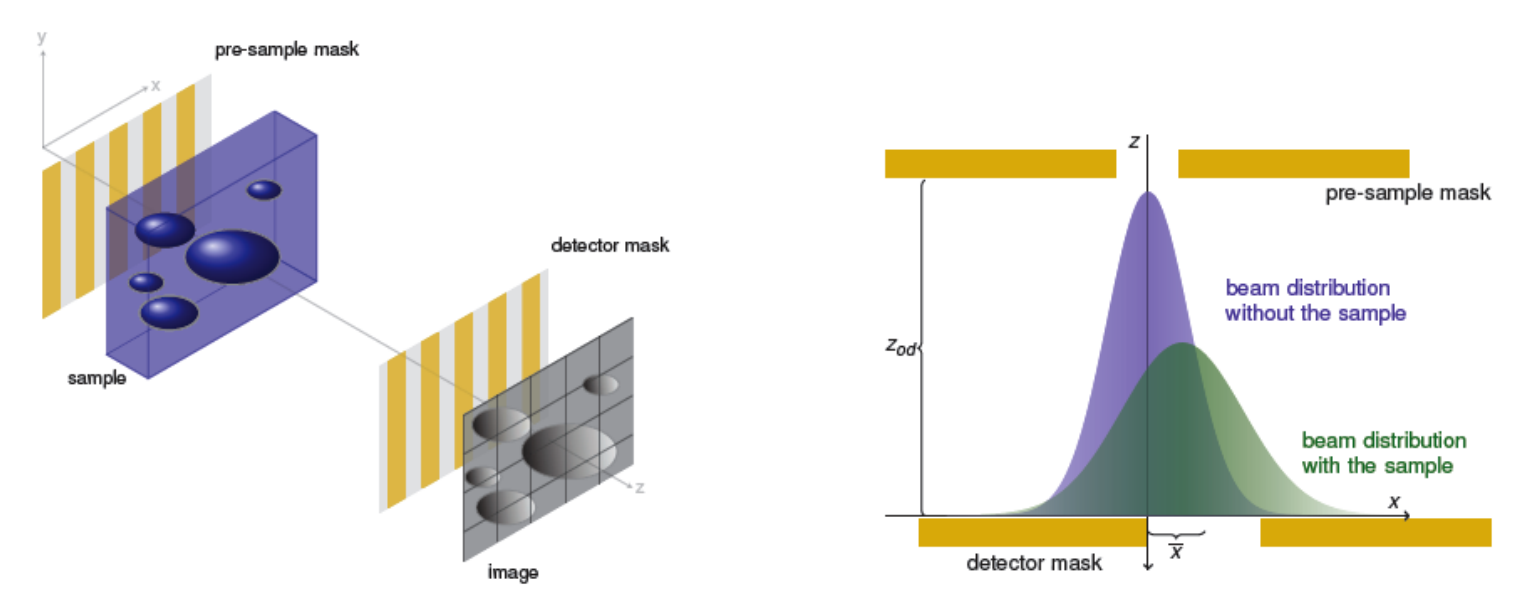
$$\Delta\theta_{x,eff} = \frac{1}{z_{od}} R^{-1}\left(\frac{I_{obj,+}}{I_{obj,-}}\right) \rightarrow \sigma\left(\Delta\theta_{x,eff}\right); \frac{\sqrt{C\left(x_{e,+}\right)}}{z_{od}\sqrt{2TI_0}\left[\rho_{ref,n}\left(x_{e,+}\right)-\rho_{ref,n}\left(x_{e,+}+d\right)\right]}$$

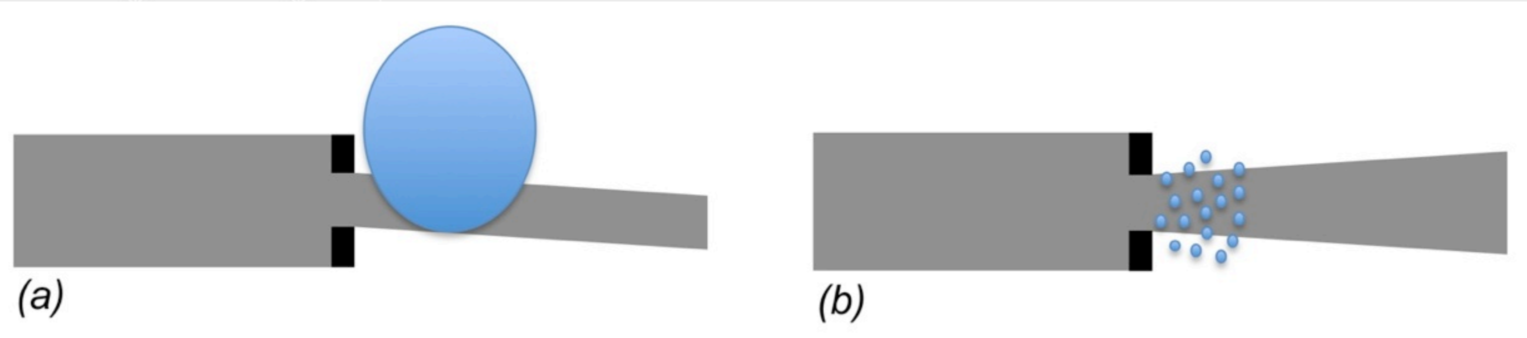
~ 250 nRad at mammographic dose

(the method can reach 2 nRad at synchrotrons with high dose)

Diemoz *et al*, Appl. Phys. Lett. 103 (2013) 244104

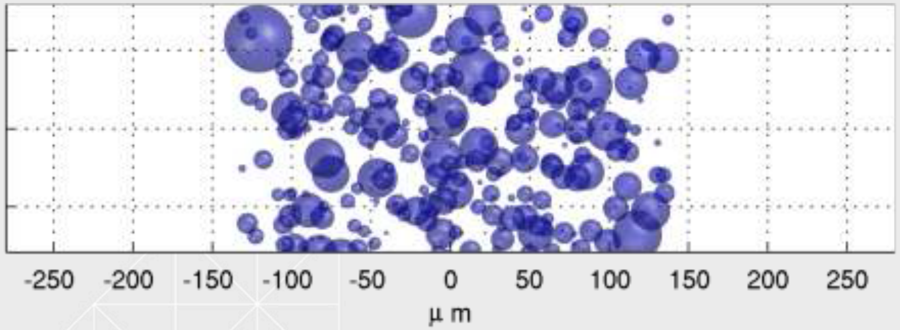
DARK FIELD (USAXS) IMAGING retrieval

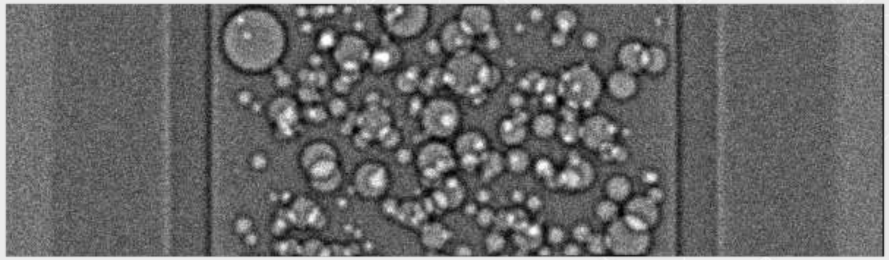




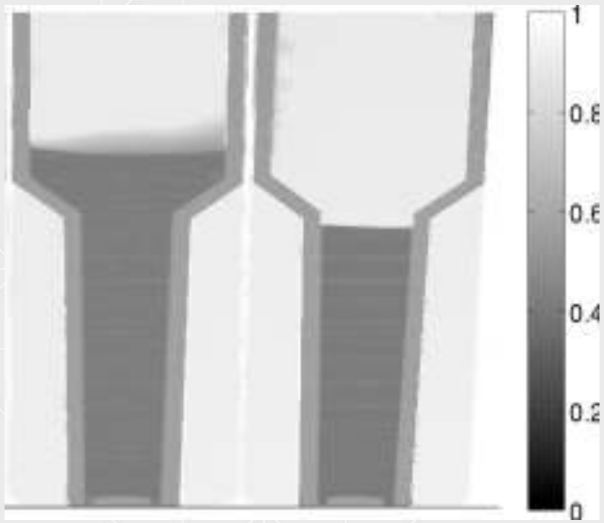
Endrizzi *et al*, Appl. Phys. Lett. 104 (2014) 024106

Microbubbles:
a new concept of "phase-based" x-ray contrast agent



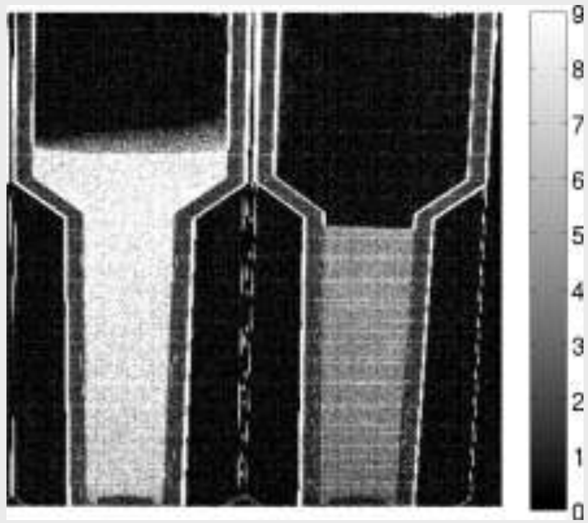


absorption



bubbles no bubbles

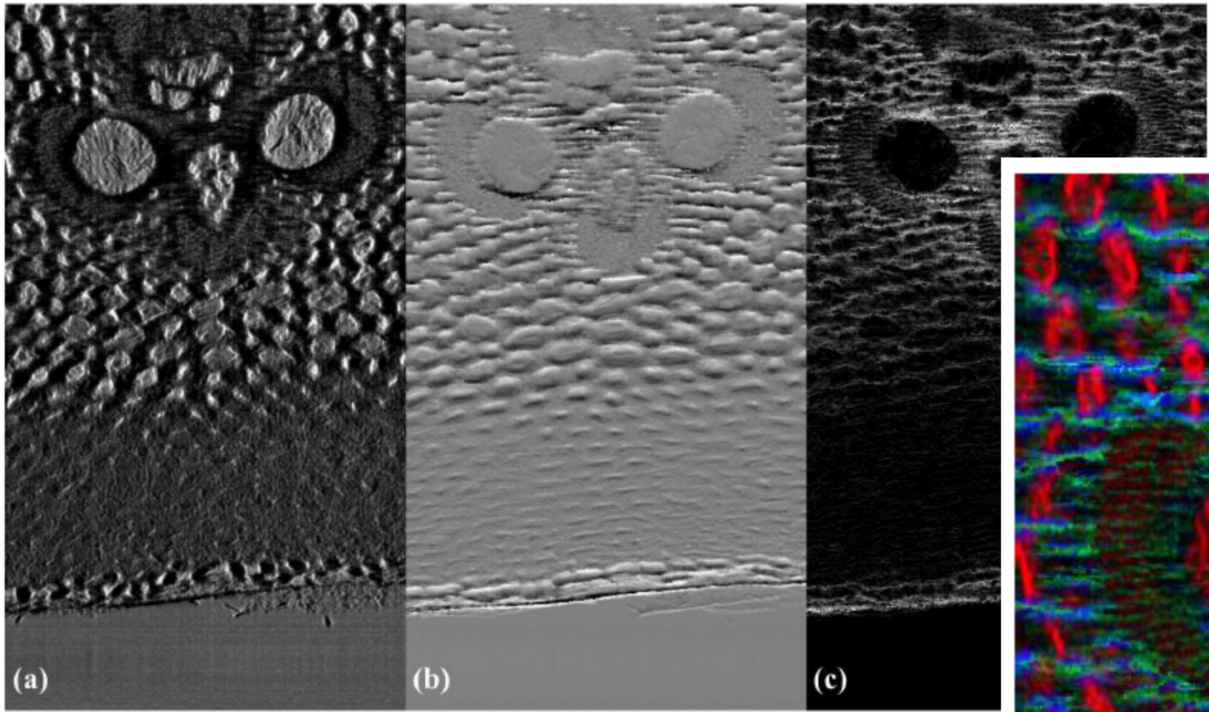
dark field

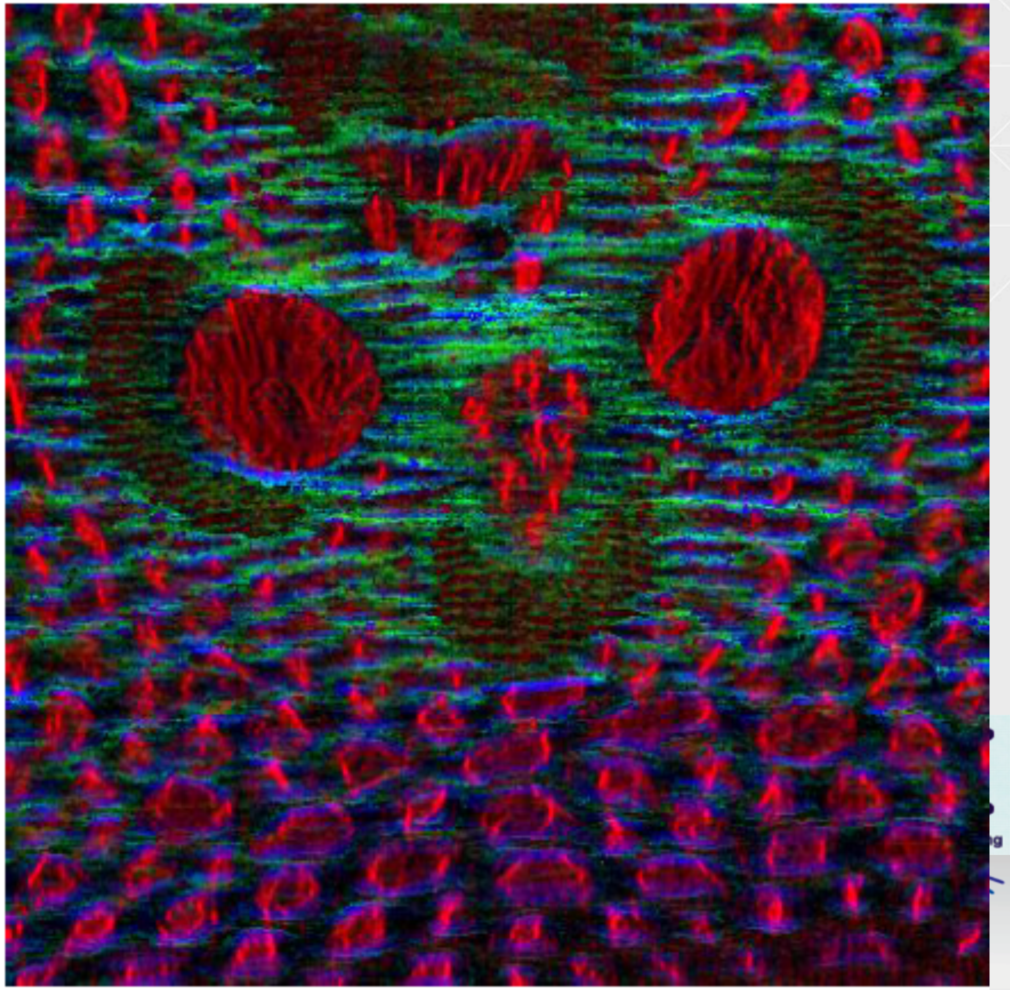


bubbles no bubbles

Millard et al. Appl. Phys. Lett. 103 (2013) 114105

Complementarity of the three signals

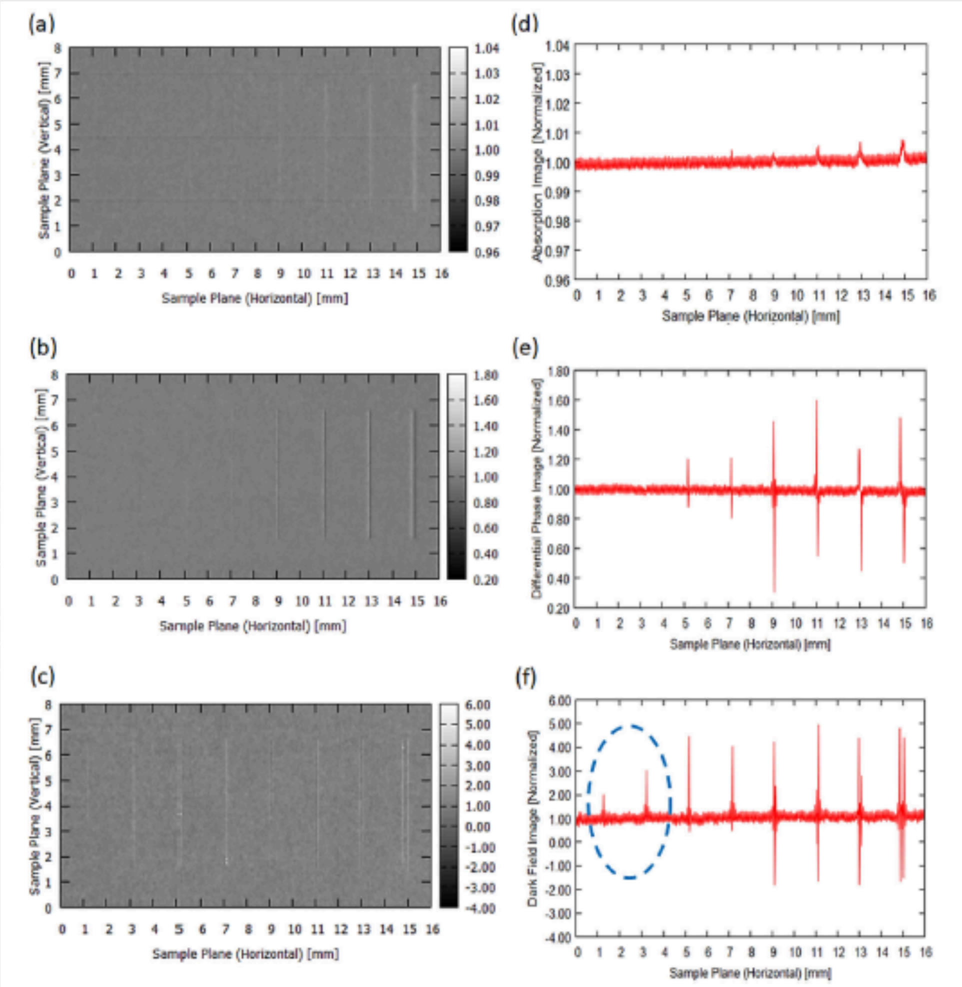


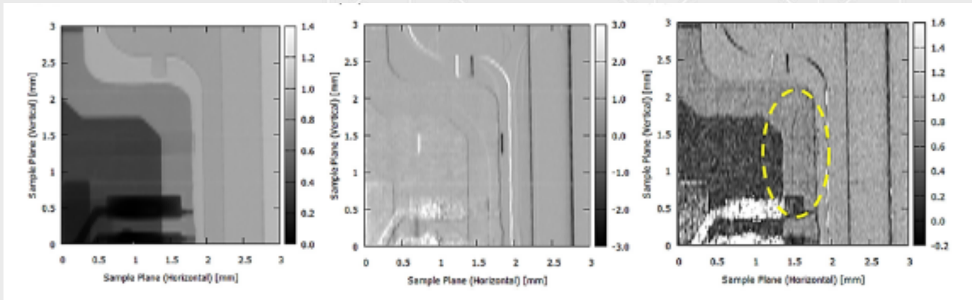


R abs
B |refr|
G scatt

Vittoria et al Appl. Phys. Lett. 104 (2014) 134102

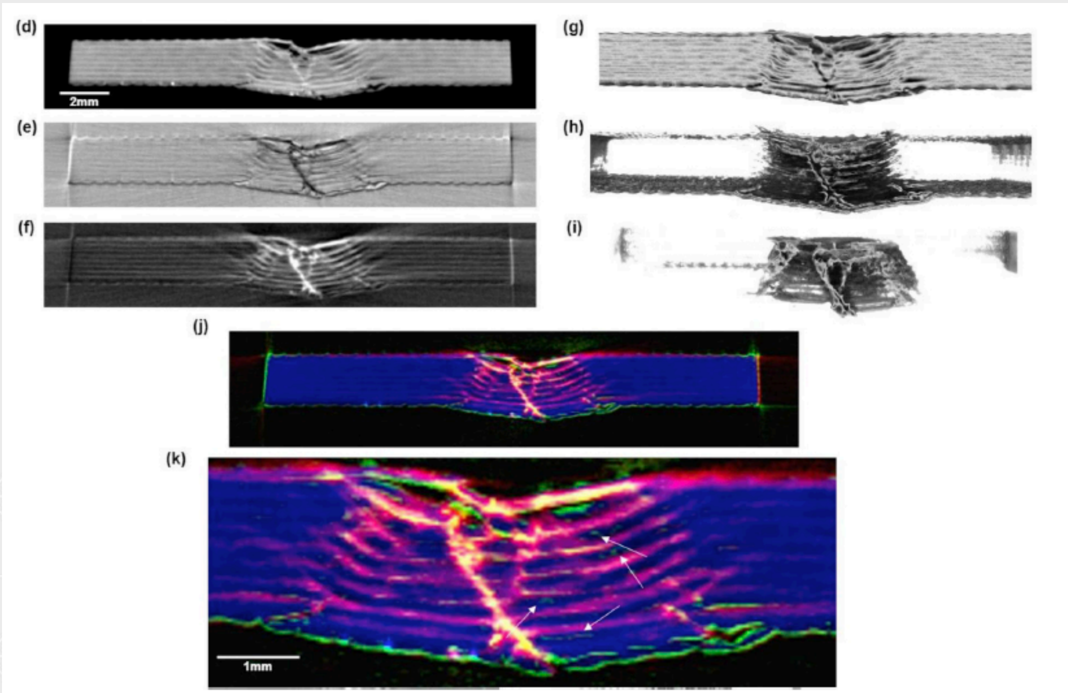
Detection of individual sub-pixel features

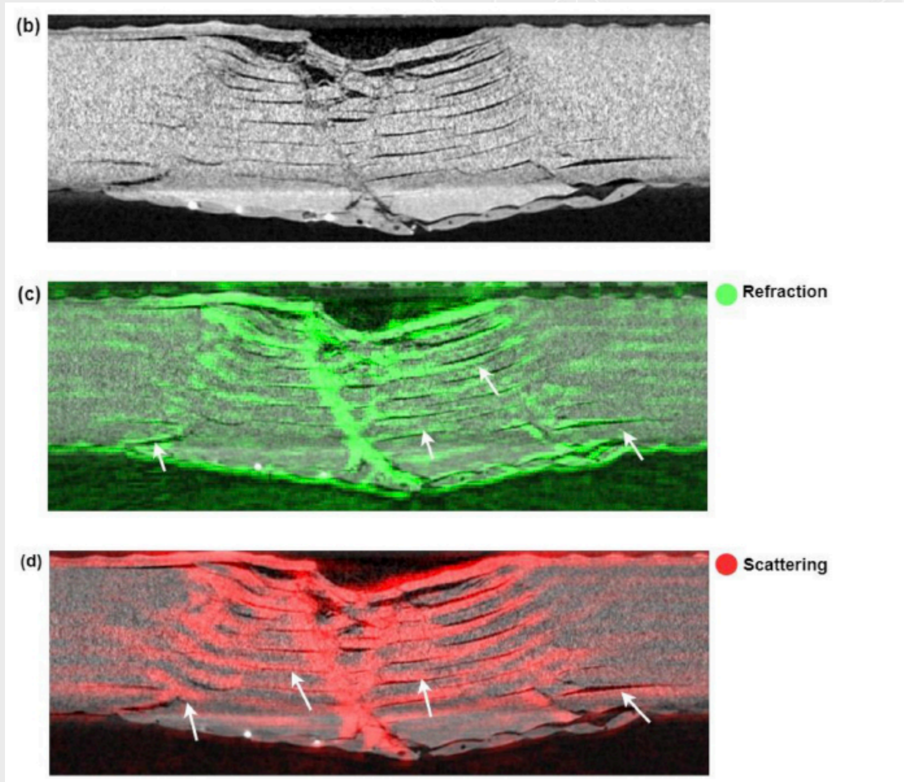




Matsunaga *et al* J. Phys. D Appl. Phys. 53 (2020) 095401

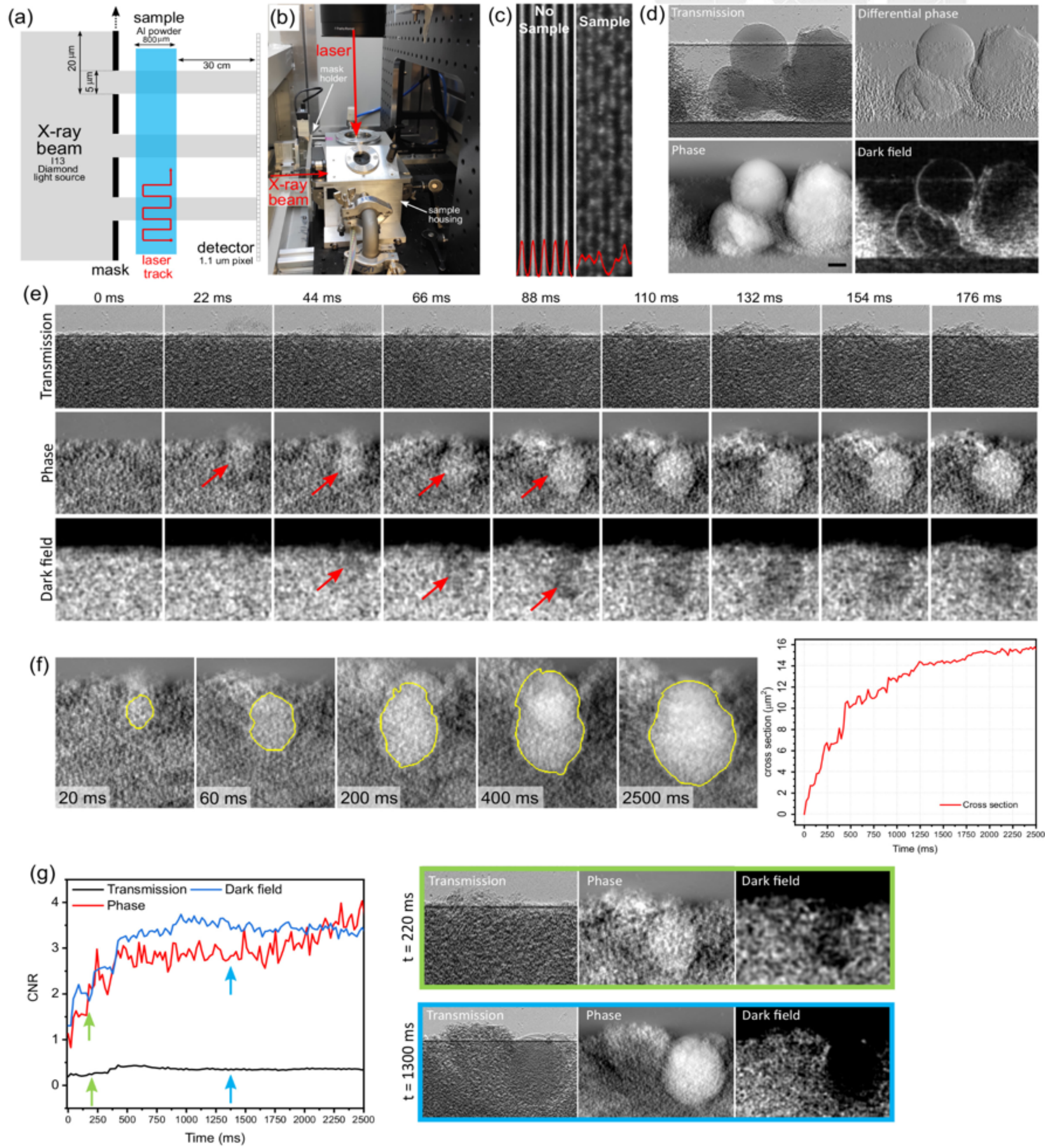
Use of complementarity on damage characterisation in composites





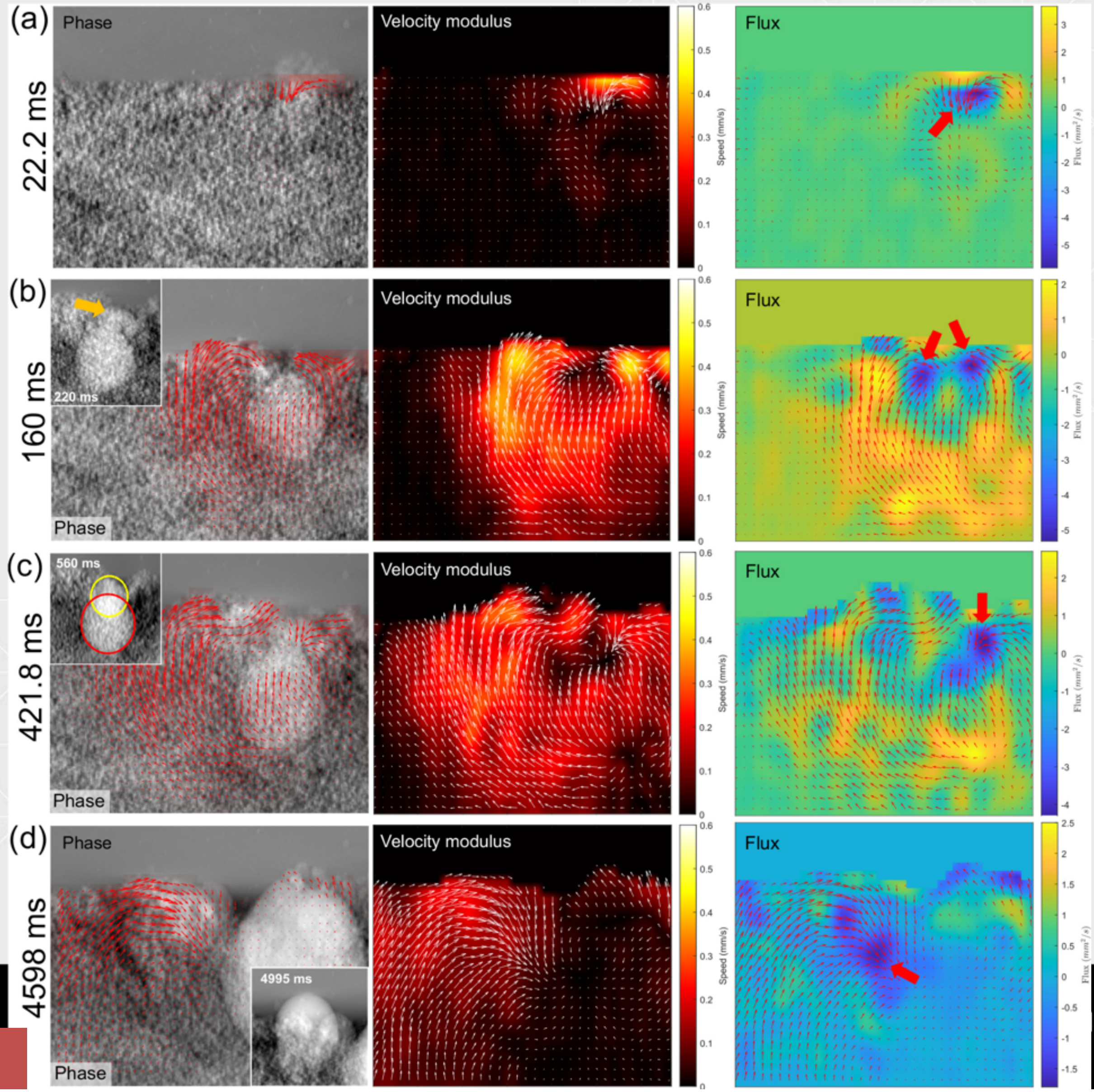
Shoukroun *et al* Compos Part B 181 (2020) 107579

POC (synchrotron) on additive manufacturing



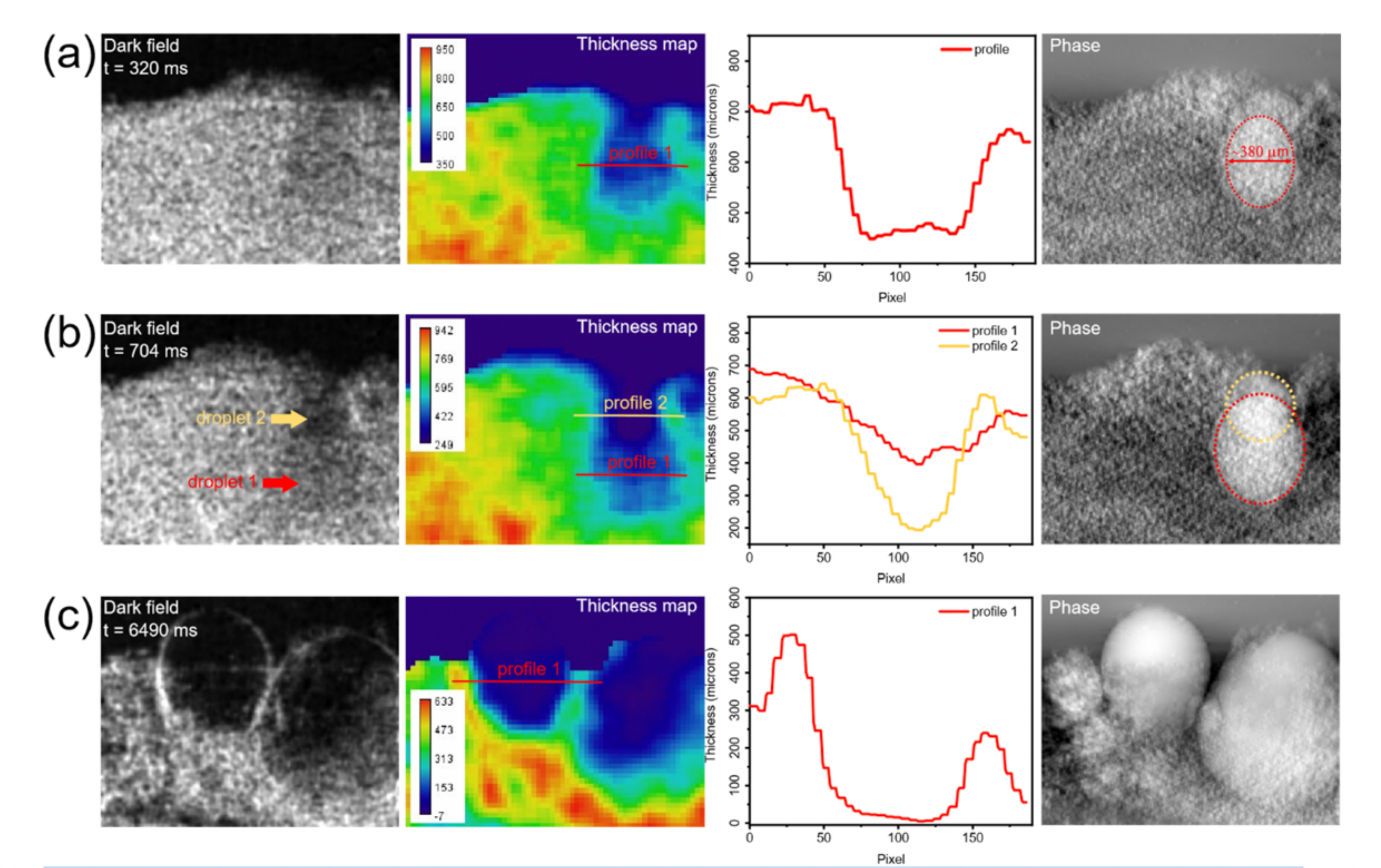
Massimi *et al*, Phys. Rev. Lett., *in press*

POC (synchrotron) on additive manufacturing



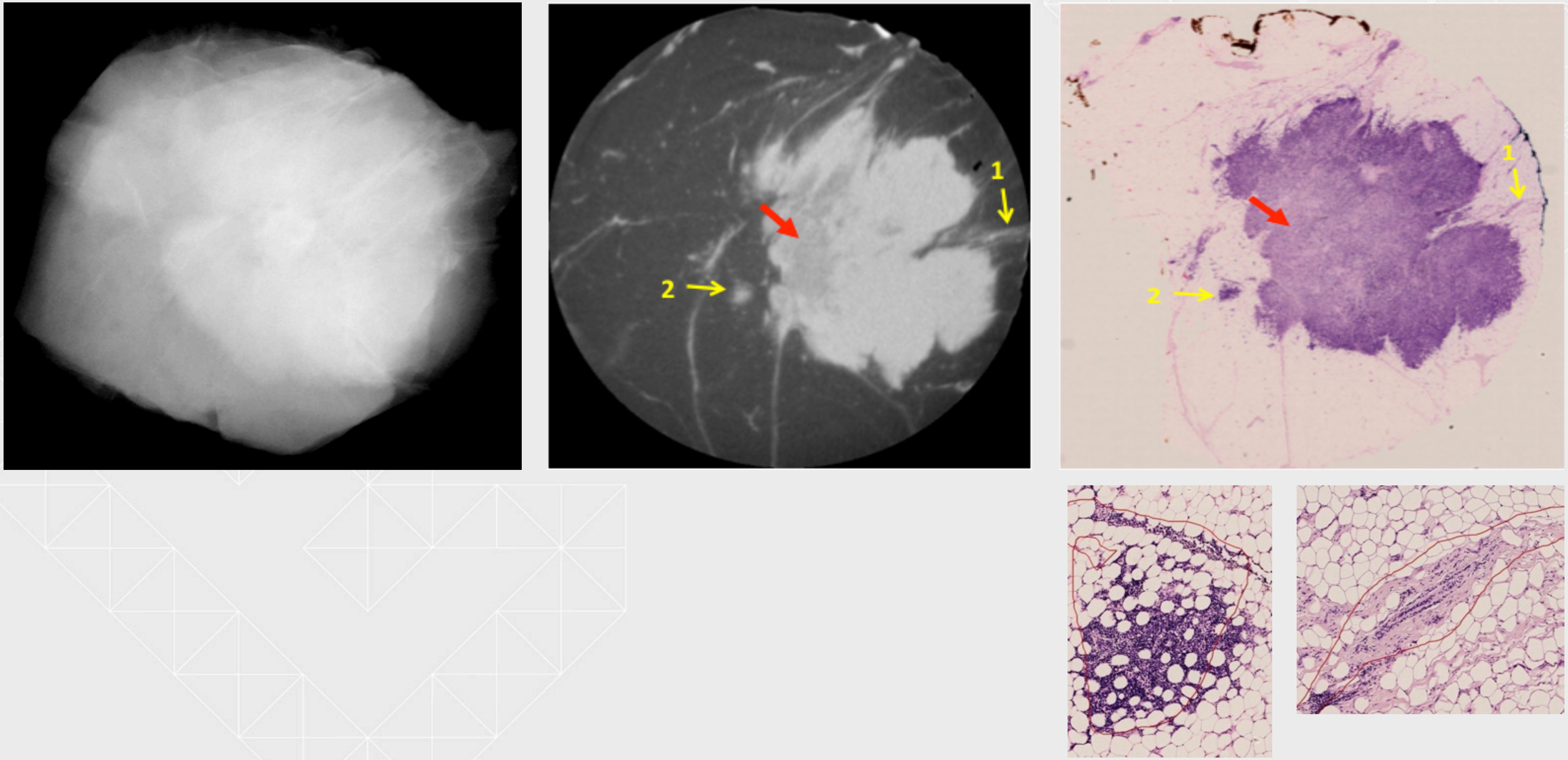
Massimi *et al*, Phys. Rev. Lett., *in press*

POC (synchrotron) on additive manufacturing



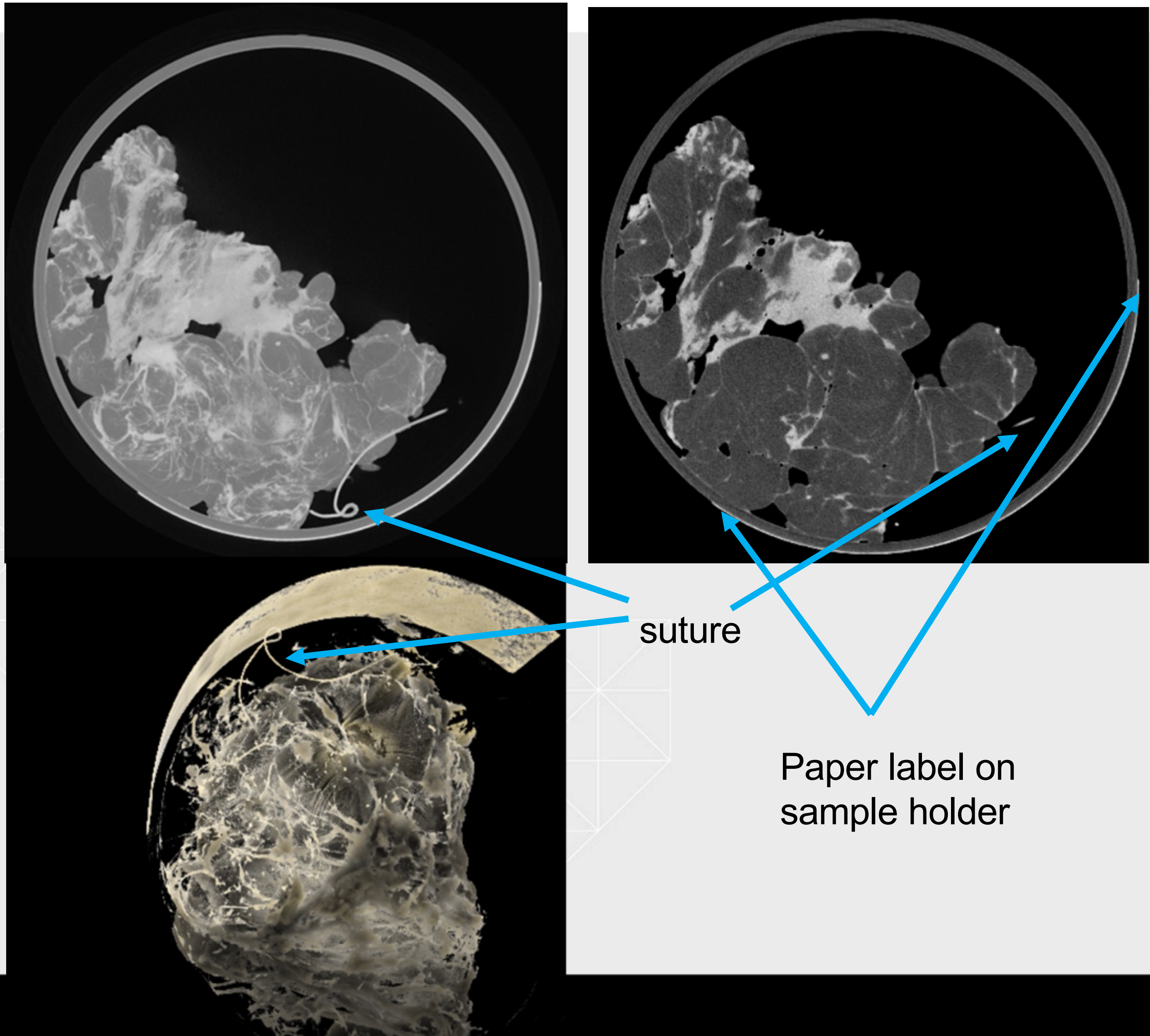
Massimi *et al*, Phys. Rev. Lett., *in press*

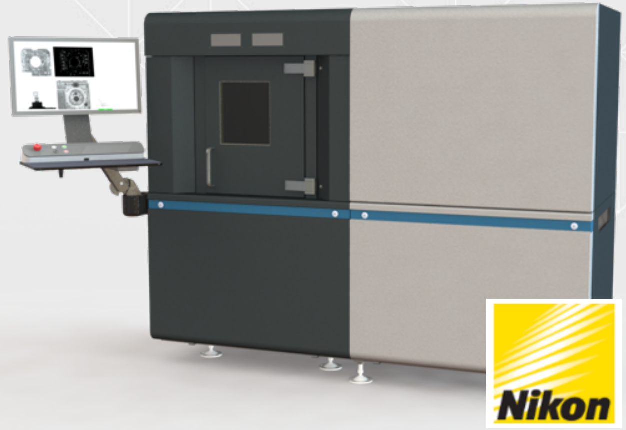
intra-operative breast imaging project



Massimi *et al* Sci Rep. 11 (2021) 3663

Real-time, large-area





Unpublished, courtesy L. Massimi and L. Jones, supported by Wellcome Trust 200137/Z/15/Z

Conclusions

Edge-illumination XPCi is non-interferometric, virtually incoherent (while remaining quantitative) - this makes it suitable for translation into real-world systems.

One key aspect is the possibility to implement single-shot methods, avoiding having to displace optical elements between acquisitions etc. We see this as absolutely essential in CT – e.g. continuous sample rotation is otherwise impossible.

By exploiting these properties, we managed to reach acquisition times compatible with real-world uses.

CT implementations are flexible both in terms of resolution and contrast modalities.

We have ramped up planar imaging to large FoVs (20 x 45 cm^2) and high x-ray energies (120-150 kVp) and the next step is to translate this into CT implementations as well.

BIG THANKS TO:

https://www.ucl.ac.uk/medical-physics-biomedical-engineering/research/groups-and-centres/advanced-x-ray-imaging-group-axim

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