Phase-Contrast X-ray imaging

Virtually all X-rays imaging in a medical or biological context is performed by measuring differences in the attenuation of X-rays by the tissues. The material properties are inferred from the differences in the number of photons detected at the end of defined paths in the sample. Naturally then this form of X-ray imaging concentrates almost entirely on the particle side of X-ray behaviour. The main atomic process responsible for the attenuation and thus forming the image is the photo-electric effect. For the purposes of this summary the photo-electric effect can be considered a binary interaction which, if it occurs, stops the X-ray, if not the X-ray continues as before. The chance of the process occurring scales very strongly with atomic number and is only linearly dependent on density and thickness. As such when imaging anatomy with reasonable differences in atomic number and density such as bone to soft tissue it can provide excellent image contrast. However for thin tissues with very similar atomic number and only small density variation (such as is often the case in soft tissue imaging) the differences in absorption rates are very small hence the resultant image contrast is generally very poor.

However X-rays also exhibit wave-like behaviour and thus when they interact with a material the interactions induce a phase-shift in the X-ray 'wave'. If the X-ray is not normally incident on the material interface this phase-shift also corresponds to an angular deflection. The deflection is generally very small (arc-seconds) and the energy transfer is very small compared to the photo electric effect. The probability of this interaction occurring can be higher by three orders of magnitude than that of the photo electric effect at typical imaging energies of a few tens of kV. However it is not trivial to measure X-ray phase since there are no direct phase reporting X-ray detectors, in common with many other imaging modalities .

There are two broad options available to detect this phase shift. The first is to arrange the system so that there are 2 or more possible X-ray paths towards a detector. The detected signal will be given by the superposition of the X-rays waves at the detector. If at least one of the X-rays traverses the sample then any induced phase-shift in this ray will affect the superposition and thus the signal as reported by the detector. There is a wide variety of geometries that can achieve this system. However they require the use of either highly coherent sources (e.g. a synchrotron) or precise optical elements to form individually spatially coherent beams. This latter method becomes extremely challenging with hard X-rays as the low wavelength of X-rays makes optical-style elements very difficult to create and align. An example geometry is shown in Figure 1, although there are many others possible[1,2].
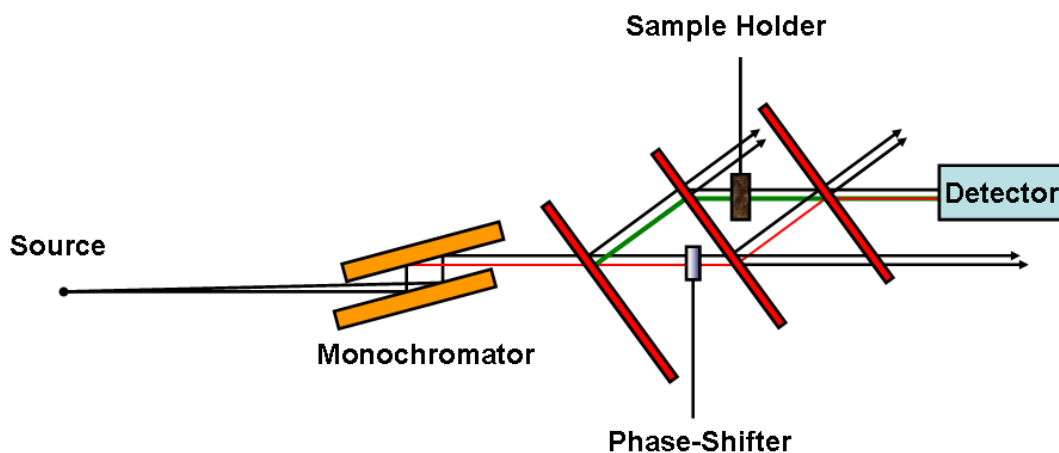

Figure 1: A possible imaging geometry which could be used to acquire phase contrast images using polychromatic X-rays, based on a monochromator and phase shifters.

The other option to assess phase is to measure the deflection of the X-rays caused by the phase change. The small deflection of the beams can be measured by using the extreme angular dependence of reflectivity that certain crystalline materials have in the X-ray spectral range. By scanning an analysing crystal over small angles whilst constantly measuring the intensity one can measure the angular spread of the radiation. By comparing this to the transmitted image, the phase can be extracted[3-6] One possible example of this geometry is shown in Figure 2.
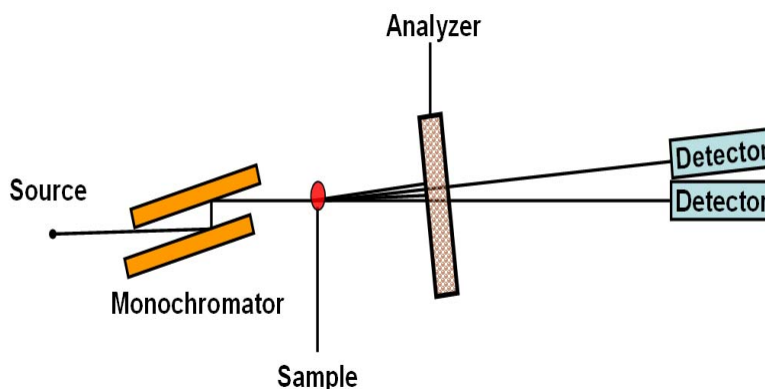

Figure 2: A possible imaging geometry which could be used to acquire phase contrast images using an analysing crystal.

Alternatively the deflection can also be inferred from the variation in intensity profile that arises from the small deflections as shown in Figure 3. The variations arise as the deflections caused by the sample push X-rays either closer together or further apart dependent on the material and the angle of incidence[7,8].
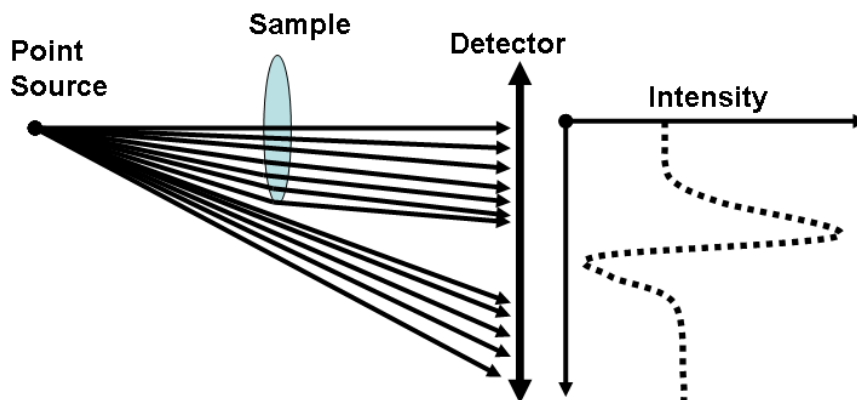

Figure 3: An alternative method of acquiring X-ray phase contrast images by exploiting a high spatial resolution detector; in practice no adequately fine spatial resolution detectors are available.

One problem with this technique is that it requires either requires very high resolution detectors or a large sample to detector distance to spread the profile out in order to observe the intensity profile.

Recently a novel approach has been presented which overcomes the shortcomings of this last measurement method. This new approach images the variation in pixel intensity due to the deflection of separate small X-ray beams, each of which is aligned to an individual pixels or a group of pixels. The smaller beams are formed by passing the much larger main X-ray beam through a

'coded' aperture. The deflection in these, now independent, small beams is then imaged by exploiting the pixel edge effect[9]. This is the effect arising when a beam passes close to or over the edge of an active detecting region. If the edge is well defined then even very small deflections in the beam vary the amount of the beam incident on the active region. As such the response of the individual pixels becomes very dependent on incidence angle. An example of this geometry is shown in Figure 4.
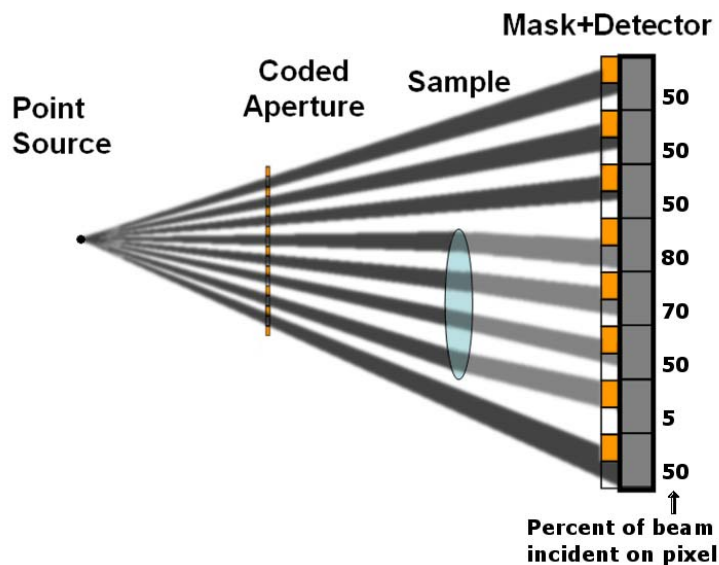
Figure 4: A practical method of acquiring phase contrast images from a polychromatic source.

A crucial advantage of this geometry[10] is that the tolerances associated with the individual X-ray beam elements and their alignment is far looser than other methods. The detector mask need only be the same pitch as the detector and the coded aperture is of the same order of magnitude for sources with small cone angles of emission.

The apparent ease of manufacture of the aperture masks and the fact that non-synchrotron sources can be used makes this a practical method which may allow us to incorporate it into the Small Animal Radiation Research Platform (SARRP, http://www.xstrahl.com/sarrp.htm). There are however some challenges: in SARRP the source to detector distance (SSD) cannot readily be increased beyond 1 m without making a rotating high precision gantry very difficult and expensive. However we are currently investigating solutions to this and if this issue can be overcome this may provide extremely low dose imaging.

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This note was prepared by A Kavanagh and B Vojnovic in October 2011