Phase contrast X-ray imaging of large samples using an incoherent laboratory source

C. Kottler*, 1, F. Pfeiffer1, O. Bunk1, C. Grünzweig1, J. Bruder1, R. Kaufmann2, L. Tlustos3, H. Walt4, I. Briod4, T. Weitkamp5, and C. David 1

1 Paul Scherrer Institut, 5232 Villigen PSI, Switzerland
2 Centre Suisse d’Electronique et de Microtechnique SA, 8048 Zurich, Switzerland
3 CERN, CERN/PH, 1211 Geneva, Switzerland
4 Universitätsspital Zürich, 8091 Zurich, Switzerland
5 Institut für Synchrotronstrahlung, Forschungszentrum Karlsruhe, 76021 Karlsruhe, Germany

Received 27 September 2006, accepted 20 February 2007
Published online 1 August 2007

PACS 07.60.Ly, 42.30.Rx, 42.87.Bg, 87.57.–s, 87.59.–e, 87.61.Ff

An interferometric method to record quantitative X-ray phase contrast images has been developed that can be used at polychromatic and incoherent X-ray sources such as laboratory tubes. With respect to previously presented results, in this work we report on recent developments and results that have been achieved in view of potential future applications such as in medicine or biology. In particular, due to improvements in the fabrication process large area diffraction gratings with high aspect ratio were achieved. Thereby, the field of view of the interferometer has been drastically increased to $64 \times 64 \text{mm}^2$ and the design value of the photon energy for the gratings could be increased up to 28 keV. Moreover, the use of a Medipix2 single photon-counting pixel detector shows a considerable improvement in image quality and sensitivity over the integrating detector used so far.

© 2007 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim

1 Introduction

The contrast of conventional X-ray images is obtained through the differences in the absorption cross section of the constituents of the object. The technique yields excellent results for highly absorbing structures, e.g., bones, embedded in a matrix of relatively weakly absorbing material, e.g., the surrounding tissue of the human body. However, when different forms of tissue with similar absorption cross-sections are under investigation (e.g., in mammography or angiography), the X-ray absorption contrast is poor. Consequently, differentiating pathologic from non-pathologic tissue from an absorption radiograph obtained with a current hospital-based X-ray system still remains very difficult for certain tissue compositions.

Besides the photo-electric absorption of X-rays inside a sample, the elastic scattering causes a phase shift of the X-ray waves. In fact this second interaction is the dominating one: if we consider 20 keV X-rays passing through a sheet of organic matter, it can be calculated that only a fraction of a percent will be absorbed, while the phase of an X-ray wave will be shifted by half a wavelength. If one could detect this phase shift, e.g. by interference with an unperturbed reference wave, the image contrast would be significantly improved.

* Corresponding author: e-mail: christian.kottler@csem.ch, Phone: +41 (0)56 310 2989
Several methods to detect the shift of X-rays passing through the sample have been investigated in the past. Excellent overviews can be found in review articles by Fitzgerald [1] and Momose [2]. The methods can be classified into crystal interferometer methods, techniques using an analyzer crystal, and free-space propagation methods. Although some of them yield excellent results for specific problems, none is very widely used. In particular, none of them has so far found medical diagnostics applications, which require a large field of view of many centimeters, the efficient use of broad-band radiation from a laboratory X-ray tube and a reasonably compact set-up.

As an alternative approach we have developed a grating based differential phase contrast (DPC) set-up [3] which can be used to retrieve quantitative phase images with polychromatic, incoherent X-ray sources [4]. The principle of the method is based on detecting minute changes in the direction of propagation, which are caused by refraction of the X-rays passing through a phase shifting object. Equivalent to refraction in the visible range, the change in direction is proportional to the local gradient in phase shift, however, it should be noted, that the refractive power of matter for X-rays is many orders of magnitude weaker. The principle of the experimental set-up is shown in Fig. 1. The essential part of the interferometer consists of two gratings placed between the object and the image detector, which act as an array of collimating slits that have a transmission depending strongly on the relative position of the two gratings and the angle of incidence. Thus, any local phase gradients in the object cause a local change in intensity recorded on the detector. While the analyzer grating close to the detector consists of an array of highly absorbing gold lines, the beam splitter grating just downstream of the object is made of phase shifting lines, which reduces the losses of the set-up. Note that the described set-up does not require monochromatic radiation. More details on the optical considerations and the data acquisition procedure can be found elsewhere [5].

![Fig. 1](image-url) a) Grating interferometer set-up using a spatially coherent X-ray source (e.g. a synchrotron radiation beamline). It consists of a beam splitter grating $G_0$, an analyzer grating $G_2$, and an image recording detector. b) The source grating $G_s$ creates an array of individually coherent but mutually incoherent line sources that provides the coherence requirements for the grating interferometer.
It is evident, that such a set-up with only two gratings (as shown in Fig. 1a) requires a source that provides sufficient spatial coherence, i.e. which is small enough and far away enough to provide a sufficiently narrow angular uncertainty of the incoming rays. Whereas this is not a problem at synchrotron, where the source is usually less than a millimeter in size and situated tens of meters away from the experiment, this poses a severe restriction for the use in laboratory equipment based on X-ray tubes, which usually cannot be placed at sufficiently large distances for reasons of required compactness and flux density. This problem can be solved by introducing a third grating just downstream of the source (see Fig. 1b), that essentially creates an array of spatially coherent line sources. If the condition \( p_0 = p_1 \times l/d \) is fulfilled, where \( p_0 \) and \( p_1 \) are the periods of the source grating and the analyzer grating, \( l \) is the source grating to beam-splitter grating distance and \( d \) is the beam-splitter grating to analyzer grating distance, then the images created by each line source are superimposed in the image plane. The implementation of such a source grating therefore makes it possible to use incoherent radiation sources, resulting in efficient use of the available flux.

The experimental demonstration of this method has previously been reported by us using a three grating set-up optimized for radiation around 17 keV photon energy [4]. The field of view was restricted to 9 mm × 9 mm by the grating fabrication technology and the image detector. In the following sections, we report on more recent experiments using gratings with areas of several square centimetres. In order to increase the penetration depth of the used radiation, we now optimized the set-up for X-ray photon energies around 28 keV. This increases the thicknesses of organic samples that can be penetrated to several centimeters, which is an important step forwards with regard to future applications such as mammography.

## 2 Experimental set-up

The gratings used in our experiments were made from Si wafers using standard photolithography techniques, anisotropic wet chemical etching and in case of \( G_0 \) and \( G_1 \) subsequent electroplating. The total size of the gratings and therefore the achievable field of view is determined as the maximum square area on the used 100 mm wafers (effective grating size: 64 mm × 64 mm). The periods of the gratings is \( p_0 = 73 \, \mu\text{m} \) for the source grating \( G_0 \), \( p_1 = 3.89 \, \mu\text{m} \) for the beam splitter grating \( G_1 \) and \( p_2 = 2 \, \mu\text{m} \) for the analyzer grating \( G_2 \). \( G_0 \) and \( G_2 \) are made by electro-deposition of gold absorber lines on silicon substrates, while \( G_1 \) consists completely of silicon. The depth of the grooves of \( G_1 \) determines the design wavelength of the interferometer because it determines at which photon energy the relative phase-shift is \( \pi \). In our set-up the depth of the grooves of \( G_1 \) was 36 \( \mu\text{m} \) which corresponds with the photon energy of 28 keV. Details on the micro-fabrication process can be found elsewhere [6]. The inter-grating distances were \( l = 1.60 \text{ m} \) (\( G_0 \) to \( G_1 \)) and \( d = 44 \text{ mm} \) (\( G_1 \) to \( G_2 \)).

Two different systems of image recording detectors can alternatively be mounted close to the analyzer grating. The first consists of a large CsI-scintillator screen (thickness 150 \( \mu\text{m} \), size: 64 mm × 64 mm), a demagnifying optical lens system and a Peltier-cooled charged coupled device (CCD) sensor. The obtained spatial resolution is \( \approx 100 \, \mu\text{m} \). The active area of this system is determined by the size of the scintillator screen and thus one can take advantage of the full size of the gratings. The second detector that can alternatively be used is one Medipix2-module [7]. The Medipix2 ASIC (Application Specific Integrated Circuit) is a high spatial, high contrast resolving CMOS pixel read-out chip working in single photon counting mode. Here, it is combined with a sensor consisting of a 700 \( \mu\text{m} \) thick absorber layer of amorphous silicon which converts the X-rays directly into detectable electric signals. The chip consists of 256 × 256 pixels with a size of 55 \( \mu\text{m} \) each. Indeed, the field of view of our Medipix2-module is limited to 14 mm × 14 mm but in comparison to our scintillator based detector it provides much better resolution and high sensitivity.

The experiments reported here were carried out on a Seifert ID-3000 X-ray generator operated at 35 kV and 30 mA. We used a tungsten line focus tube (DX-W8x0.4-L) with an effective source size of 0.8 mm × 0.4 mm (horizontal × vertical). Typical exposure times per sample were on the order of a few minutes. This value could be reduced 2 orders of magnitude by using a rotating anode source.
3 Results and discussion

In Figs. 2–4 examples of images of some selected samples are displayed that were acquired with our set-up. Consider first the example of the chicken wing shown in the photograph in Fig. 2a that was measured as prepared for alimentary purposes. Whereas the conventional absorption radiograph (Fig. 2b) mostly shows the structure of the strongly absorbing bones, in the differential phase contrast image (Fig. 2c) the soft tissue and bones are clearly visible.

Fig. 2 (online colour at: www.pss-a.com) Photograph (a), absorption image (b) and differential phase contrast image (c) of a chicken wing.

Fig. 3 Comparison of image quality between combined scintillator/CCD-detector (a, b) and Medipix2 single photon counting detector (c, d) shown at the example of two conches for both absorption and phase gradient image. The spatial resolution reached with the Medipix2 detector is 55 µm whereas the resolution obtained with the scintillator/CCD system is ≈100 µm.
Fig. 4 Absorption (a), local phase gradient (b) and integrated phase image (c) of the top part of a dried hornet recorded with the Medipix2 detector.

(Fig. 2c) more details are revealed of the surrounding soft tissue. In particular, the wing’s outer contour is very pronounced since the difference in index of refraction of the tissue and the air gives rise to a steep phase gradient and therefore a sharp contrast. Furthermore, the structure of the tendon can be clearly recognized in Fig. 2c (right edge of the wing’s contour).

Figure 3 shows the results obtained from two conchés, one recorded with the large area scintillator based detector (Fig. 3a and b, image size: 64 × 64 mm²) and the second recorded with the Medipix2-module (Fig. 3c and d, image size: 14 × 14 mm²). Apparently, the data set acquired with Medipix2 detector shows the internal structure of the conch in much more detail than the one with the scintillator based detector which has fairly moderate spatial resolution.

The measurement of the top part of a previously dried hornet obtained with the Medipix2 detector is shown in Fig. 4. Both images, absorption (Fig. 4a) and phase gradient (Fig. 4b), nicely show the details of the hornets chitin constituents such as its legs that can be seen on the left hand side in both of these images. In Fig. 4c the result is shown that is trivially obtained integrating the differential phase contrast image along the direction of the phase gradient (horizontal direction). Despite of some artifacts that arise upon integration it gives a quantitative phase image of the wave-front profile with satisfactory quality for quantitative analysis.

4 Conclusion

The measurements reported here show that X-ray grating interferometry is a technique that can be applied to retrieve images of local phase gradient of large samples (several centimeters). From the point of view of grating manufacturing it is feasible to produce gratings on larger silicon wafers than currently on 100 mm wafers. Thus, even a larger field of view may be realized in future. For typical applications in X-ray imaging the realistic samples may not only become larger but also thicker. Therefore, it is an important step to produce gratings with high aspect ratio and thus move the design wavelength towards higher energies. In our measurements the interferometer was designed for 28 keV photon energy. This allows for measurement of organic samples with thickness up to a few centimeters without suffering from beam hardening. As the example of the hornet in Fig. 4 shows our technique provides fully quantitative images of the wave-front phase profile which is an important prerequisite to reconstruct volumetric information of refractive index from tomographic data. Finally, we conclude that with respect to the previously presented work [4] our results here mean a major step forward in view of the feasibility of this technique for a medical, biological or any other X-raying application that might benefit from phase contrast.

Acknowledgements This work was supported by the Swiss Commission for Technology and Innovation KTI/CTI under contract 7796.2 DCPP-NM.
References