

# X-ray phase-contrast imaging

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## Abstract

X-ray imaging is a standard tool for the non-destructive inspection of the internal structure of samples. It finds application in a vast diversity of fields: medicine, biology, many engineering disciplines, palaeontology and earth sciences are just few examples. The fundamental principle underpinning the image formation have remained the same for over a century: the X-rays traversing the sample are subjected to different amount of absorption in different parts of the sample. By means of phase-sensitive techniques it is possible to generate contrast also in relation to the phase shifts imparted by the sample and to extend the capabilities of X-ray imaging to those details that lack enough absorption contrast to be visualised in conventional radiography. A general overview of X-ray phase contrast imaging techniques is presented in this review, along with more recent advances in this fast evolving field and some examples of applications.

*Keywords:* X-ray; phase-contrast; imaging

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## 1. Introduction

2 The use of X-rays for imaging the internal structure of samples quickly spread  
3 around the world soon after the first X-ray radiograph was taken by Wilhelm  
4 Conrad Röntgen towards the end of 1895 [1]. Great improvements have con-  
5 stantly been made throughout the last century both with regard to the X-ray  
6 generators and to the image receptors, including transformative advances such

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7 as the introduction of tomography [2]. X-ray imaging is nowadays a standard  
8 tool in many diverse fields and disciplines, ranging from medical sciences to ma-  
9 terials engineering and including quality control in industry as well as security  
10 screening.

11 Despite tremendous progress, the fundamental working principle has re-  
12 mained unchanged for over a century: contrast is generated by differences in the  
13 absorption of the X-rays within the sample. This can provide excellent results  
14 when relatively high attenuation exists, but leads to poor image quality when the  
15 sample is weakly absorbing. Generally speaking, this occurs for materials and  
16 tissues composed of light elements. The possibility of performing phase-based  
17 imaging bears the potential of making visible what would be undetectable with  
18 the conventional method for these classes of samples.

19 A number of reviews is already available on this topic, including a focus  
20 on the evolution and relative merits of these imaging techniques [3], on the  
21 transition from synchrotron to conventional sources [4], on medical applications  
22 [5–7] with the translation towards clinical implementation [8] the imaging of the  
23 breast [9], and also on materials science applications [10, 11].

24 The aim of this review is to present a general overview of the essentials X-  
25 ray phase-contrast imaging techniques in the hard X-ray regime, as well as some  
26 examples of use in applied investigations. An in-depth discussion is dedicated  
27 to the principles and recent advances of edge illumination, a technique that has  
28 been intensively investigated in the recent time by our group for the translation  
29 of these advanced X-ray imaging techniques into table-top instrumentation that  
30 can be compatible with clinical or industrial environments.

## 31 **2. Methods**

32 X-ray imaging is a general term that embraces an extremely wide set of tech-  
33 niques that are used to produce a representation of the sample under inspection.  
34 In order to describe phase-contrast X-ray imaging techniques, we will start from  
35 the basis of the more conventional, or absorption-based, approach.

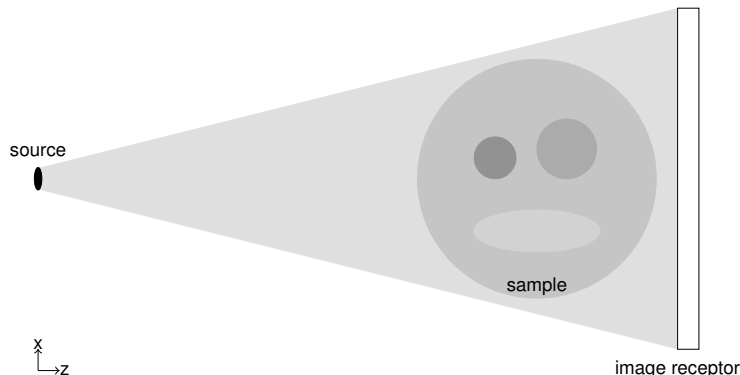


Figure 1: Schematic of the set-up for conventional X-ray imaging: the image receptor looks directly at the radiation source and through the sample.

36 *2.1. Absorption imaging*

37 The sketch in Figure 1 reports the arrangement that is typically used in ra-  
 38 diography by using an X-ray source and an image receptor. It is a transmission-  
 39 type imaging modality in which the image receptor looks at the source and  
 40 through the sample. The internal structure of the sample can be inspected in  
 41 this way because the differences in the attenuation of the X-rays, along their  
 42 trajectories from the generator to the receptor, produce contrast. In order to  
 43 quantify this effect we will use a two dimensional representation of a simple  
 44 object, a sphere made of a single material embedded into another homoge-  
 45 neous material. This situation is depicted in Figure 2 where panel 2a shows  
 46 the arrangements of X-ray radiation, phantom and detector while the panel 2b  
 47 the resulting image. The coordinate system is defined as follows: the X-rays  
 48 propagate from the source along the  $z$  axis, the object extends in all the three  
 49 dimensions and the image at the receptor is a two dimensional distribution of  
 50 intensity in the  $(x, y)$  plane. By using monochromatic radiation of wavelength  
 51  $\lambda$ , the intensity at the detector can be described by using the Beer-Lambert law  
 52 [13]

$$I(x, y) = I_0(x, y) \exp [-(\mu_o^\lambda - \mu_h^\lambda)T_o(x, y)] \quad (1)$$

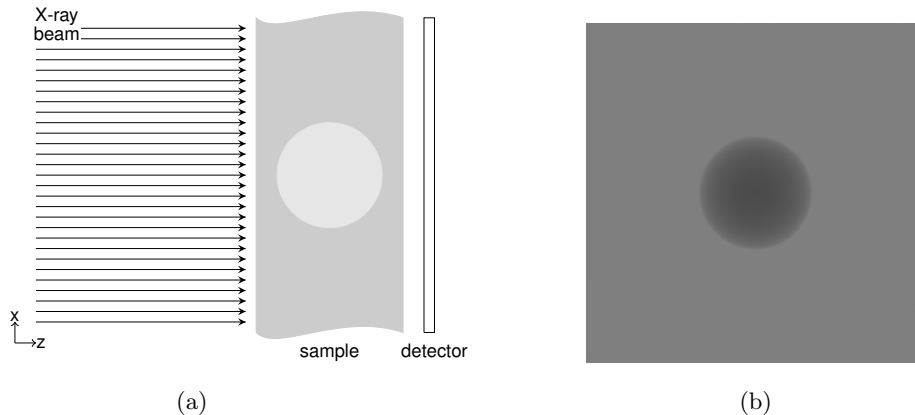


Figure 2: (a) simple model for the generation of contrast in absorption-based X-ray imaging. (b) corresponding image recorded represented as the two dimensional distribution of intensity, which was computed by using the X-Tract software [12].

53 where  $T_o(x, y)$  is the projected thickness of the object on the  $(x, y)$  plane and  
 54  $I_0(x, y)$  is the intensity incident on the sample. For the simple case presented  
 55 above this can be calculated analytically. Let us assume a sphere of radius  $r$   
 56 and centred in the origin. The projected thickness of the sphere, measured by a  
 57 line profile running across the centre of the sphere ( $y = 0$ ), is given by the real  
 58 part of

$$T(x) = 2\sqrt{r^2 - x^2}. \quad (2)$$

59 We can then calculate the corresponding intensity profile by using Equation 1.  
 60 In order to do so we need to specify the working energy (30 keV), the materials  
 61 (aluminium for the sphere and water for the embedding material) and we further  
 62 assume a constant incident intensity  $I_0(x, y) = 1$ .

63 It is often the case, for example when using conventional laboratory sources  
 64 such as X-ray tubes, that the radiation is polychromatic and its spectrum ex-  
 65 tends over a range of several tens of keV. This can be included in Equation 1 by  
 66 an integration over the energy that takes into account the energy dependence of  
 67 the source spectrum  $I_0(\lambda)$ , of the attenuation coefficient  $\mu^\lambda$  and of the detector

68 response  $\mathcal{D}(\lambda)$

$$I(x, y) = \int d\lambda I_0(x, y; \lambda) \exp [-(\mu_o^\lambda - \mu_h^\lambda)T_o(x, y)] \mathcal{D}(\lambda). \quad (3)$$

69 Each monochromatic component of the X-ray beam contributes independently  
70 to the contrast, with a weight that is equal to the relative probability of emis-  
71 sion and detection, and with the attenuation coefficient characteristic of that  
72 particular energy (for example see [14]).

### 73 2.2. Phase-contrast imaging

74 The phase of the waves travelling through the sample contributes to the  
75 modulation of the detected intensity in an X-ray phase-contrast imaging system.  
76 This can be described by means of the complex refractive index [15]

$$n = 1 - \delta + i\beta \quad (4)$$

77 where the decrement to unity  $\delta$  governs the phase shifts while  $\beta$  the absorption.  
78 Away from absorption edges, and in the region where the photoelectric effect  
79 dominates absorption,  $\delta$  and  $\beta$  can be expressed as functions of the electron  
80 density  $\rho$  and of the radiation wavelength in the following way [16]

$$\delta(\lambda) = \rho \frac{r_e \lambda^2}{2\pi} \quad (5)$$

$$\beta(\lambda) = \mu(\lambda) \frac{\lambda}{4\pi} \quad (6)$$

81 where  $r_e$  is the classical electron radius. It is worth noting that  $\delta$  is typically  
82 larger than  $\beta$ . By taking for example water at 30 keV, we obtain  $\delta \approx 2.56 \cdot 10^{-7}$   
83 and  $\beta \approx 1.36 \cdot 10^{-10}$ . Another key difference between the two parameters is their  
84 dependence on the X-ray energy  $E$ :  $\beta$  decreases approximately with  $E^{-4}$  while  
85  $\delta$  approximately as  $E^{-2}$ . Real and imaginary parts of the complex refractive  
86 indices of two materials, one composed of light elements and one composed of  
87 heavier elements, are plotted in Figure 3 in the energy range between 10 and  
88 120 keV for illustration purposes.

89 The phase shift imparted by the sample to the X-ray wave is given by

$$\Phi(x, y; \lambda) = -k \int_{\mathcal{O}} dz \delta(x, y, z; \lambda) \quad (7)$$

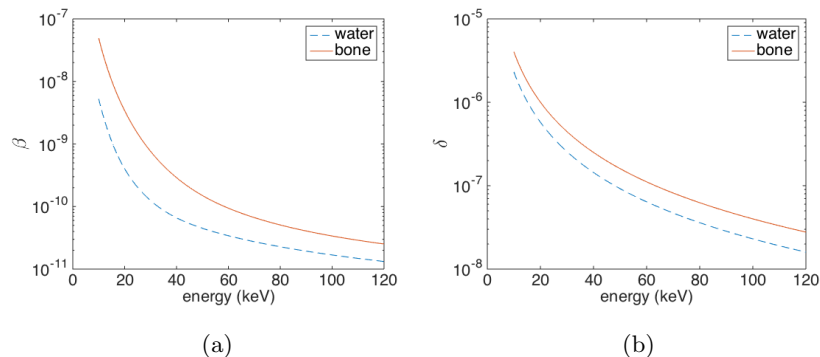


Figure 3: Complex index of refraction for water and bone as a function of X-ray energy: (a)  $\beta$  and (b)  $\delta$ . Bone composition was taken from the ICRU report [17].

90 where the integration is carried out over the extent of the object  $\mathcal{O}$  along the  
 91 optical axis, and this Equation can be considered valid for propagation through  
 92 thin objects, for which the projection approximation holds [15].

93 Referring to previous example of a simple sphere composed of a single ma-  
 94 terial, the transmission and phase shift of the sample become

$$I(x, y; \lambda) = \exp \left[ -\frac{4\pi\beta(\lambda)}{\lambda} T(x, y) \right] \quad (8)$$

$$\phi(x, y; \lambda) = -k \delta T(x, y) \quad (9)$$

95 where a single energy was used for the X-ray beam.

### 96 3. X-ray phase-contrast imaging

97 It is not possible to directly measure the phase of electromagnetic waves  
 98 at optical frequencies and above, however, phase effects can play a significant  
 99 role in the image formation also in the hard X-ray regime. Phase-contrast  
 100 imaging techniques exploit the phase perturbations introduced by the sample  
 101 to modulate the intensity recorded at the image receptor, in such a way that  
 102 these effects can be detected and interpreted.

103 A summary of these techniques will be presented in the following sections.  
 104 The classification is inevitably made afterwards, and it is therefore natural that

105 the categories will be appropriate in certain cases while less accurate in others.  
 106 X-ray phase-contrast imaging techniques are evolving fast, and a large degree  
 107 of contamination often exists across different approaches. A classification based  
 108 on the most prominent characteristics of the experimental set-ups and their  
 109 working principles was chosen here as the main criterion for distinction between  
 110 different approaches.

### 111 3.1. Interferometry

112 The first example of X-ray phase-contrast imaging method is the X-ray in-  
 113 terferometer [18, 19] which was built from a monolithic crystal and used a Laue-  
 114 Laue configuration. A schematic representation of this device is shown in  
 Figure 4. Phase-coherent beams are formed by dividing the incoming X-ray

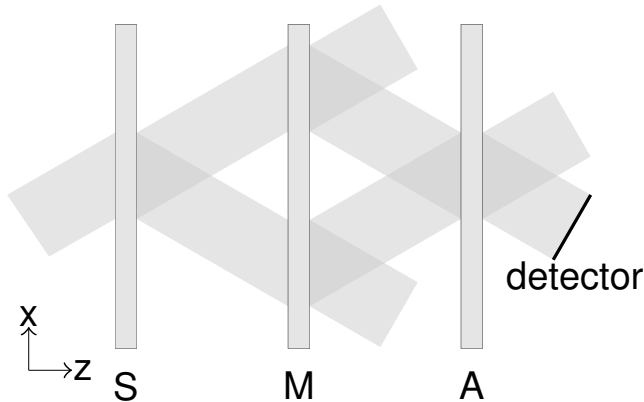


Figure 4: Top view of an X-ray interferometer.

115  
 116 beam at the beam splitter S and successively at the transmission element M,  
 117 they meet again at the analyser A where an atomic-scale standing wavefield is  
 118 formed [20]. In an ideal scenario, where the wave is perfectly planar and the  
 119 crystal free of imperfections, the field would be perfectly uniform until a sample  
 120 is introduced in one of the arms of the interferometer. The image would then  
 121 record the phase changes induced by sample, modulo  $2\pi$ . In practice, local phase  
 122 shifts, arising for example from strain and defects in the crystal, will generate  
 123 interference patterns that will be superimposed to the modulations imposed by

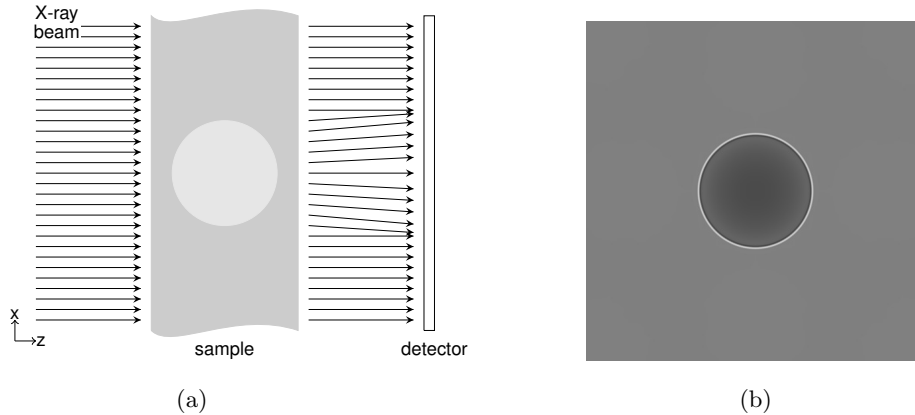


Figure 5: (a) simple model for the generation of a phase contrast image with the free space propagation technique. (b) corresponding image recorded represented as the two dimensional distribution of intensity, calculated by using the X-Tract software [12].

124 the sample. It is also possible to insert a known phase modulation in one arm,  
 125 as for example a linear phase ramp imposed by using a wedge, while the sample  
 126 under study is placed in the other arm. The linear phase ramp will generate a  
 127 series of linear fringes in the intensity recorded at the detector, hence the name  
 128 of fringe scanning method [21].

129 Following the first demonstration of the working principle and pioneering  
 130 imaging experiments [22, 23], this method was used for biomedical imaging  
 131 experiments [24–27] to study different tissue types such as breast, brain and  
 132 blood [28–30].

### 133 3.2. Free-space propagation

134 Free-space propagation techniques are perhaps the ones requiring the sim-  
 135 plest set-up because the introduction of an appropriate propagation distance  $R_2$   
 136 between the sample and the image receptor can be sufficient to make phase ef-  
 137 fects detectable (see Figure 5). Early works demonstrating this possibility date  
 138 back to the mid '90s and used both monochromatic and collimated synchrotron  
 139 radiation [31, 32] and polychromatic radiation from a microfocus X-ray tube  
 140 [33]. This phenomenon can be interpreted in terms of Fresnel diffraction and



141 key features of this approach to imaging can be identified by referring to the  
 142 following expression [15, 33, 34]

$$I(x, y; M, \lambda) = \frac{I_0}{M^2} \left[ 1 + \frac{R_2 \lambda}{M 2\pi} \nabla_{\perp}^2 \phi(x, y; R_1, \lambda) \right] \quad (10)$$

143 that describes the intensity distribution at the image receptor plane from a pure  
 144 phase object.  $M = (R_1 + R_2)/R_1$  is the geometrical magnification. The contrast  
 145 from a pure phase object vanishes when  $R_2 \rightarrow 0$ , which is the typical condition  
 146 for conventional (contact) radiography and the phase term is directly propor-  
 147 tional to the propagation distance  $R_2$ . Another feature of interest is that the  
 148 monochromaticity of the radiation is not essential for this type of imaging. A  
 149 necessary condition, however, is that the radiation must have a certain degree  
 150 of spatial coherence [13]:

$$l_c = \frac{\lambda R_1}{\sigma_s 2\sqrt{2 \log 2}} \quad (11)$$

151 where  $\sigma_s$  is the standard deviation of the source intensity distribution. The  
 152 coherence length  $l_c$  has to be comparable to or larger than the inverse spatial  
 153 frequency of the feature of interest [35] in order to obtain significant phase con-  
 154 trast. In practice this means that the source has to be relatively small or that  
 155 the object must be placed at a relatively large distance  $R_1$  from it. Another re-  
 156 quirement is that the imaging system must have spatial resolution high enough  
 157 to not wash out the interference fringes. This is conveniently summarised by  
 158 the following expression [36, 37]

$$\sigma_t^2 \approx \left( 1 - \frac{1}{M} \right)^2 \sigma_s^2 + \frac{\sigma_d^2}{M^2} + \sigma_m^2 \quad (12)$$

159 where  $\sigma_t$  and  $\sigma_d$  are the standard deviations of the system's and of the detec-  
 160 tor's point spread function, respectively. Another point to be noted is that the  
 161 diffraction term

$$\sigma_m = \frac{1}{2} \sqrt{\frac{\lambda R_2}{2}} \quad (13)$$

162 becomes less significant for increasing X-ray energies.

163 The intensity projection image, acquired with a certain propagation distance  
 164 between the sample and the detector, will contain a mixture of contributions

165 from both the absorption and the phase shifts in the sample. Other experimental  
166 parameters, like the X-ray energy, the geometrical magnification, the radiation  
167 coherence and the system resolution, determine the modulation of intensity at  
168 the detector. The process that aims at making this type of imaging quantitative  
169 by calculating phase and amplitude at the exit surface of the sample is called  
170 phase retrieval. Methods to achieve this, in the case of non-interferometric hard  
171 X-ray imaging techniques, started developing soon after the first experiments  
172 [38–41], also including polychromatic X-ray beams [42] and even exploiting dif-  
173 ferent energies for the phase retrieval process itself [43]. Quantitative retrieval  
174 algorithms are also a fundamental component for accurate three-dimensional  
175 reconstructions [44–46]. In general terms, the determination of both amplitude  
176 and phase requires more than a single measurement (for example by changing  
177 the propagation distance or by changing the energy) unless some constraints can  
178 be imposed on the sample. Quantitative phase retrieval can be performed from  
179 a single defocus distance by requiring homogeneity of the sample [47]: although  
180 this might not always be strictly satisfied, it is a very reasonable approxima-  
181 tion in many cases of interest (e.g. soft tissue samples) and this approach often  
182 delivers high quality projections and reconstructions.

183 Applications of free-space propagation X-ray phase-contrast imaging are vast  
184 and definitely too many to be covered here. We will limit this discussion to few  
185 highlights like micro- and nano-tomography applications [48–54], lung imaging  
186 [55–59] and breast tissue imaging [60, 61] including in-vivo [62–64].

### 187 *3.3. Analyser based imaging*

188 Analyser-based methods make use of crystals both for beam preparation and  
189 analysis. The crystal arrangement preceding the sample is used to monochro-  
190 matise and collimate the incoming X-ray beam while the one preceding the  
191 detector serves as a fine angular filter. A typical synchrotron experimental set-  
192 up is sketched in Figure 6. The X-ray beam is usually wide enough to cover  
193 the extent of the sample along the  $x$  direction while scanning along  $y$  is often  
194 necessary to build a two-dimensional image. The intensity at the detector is

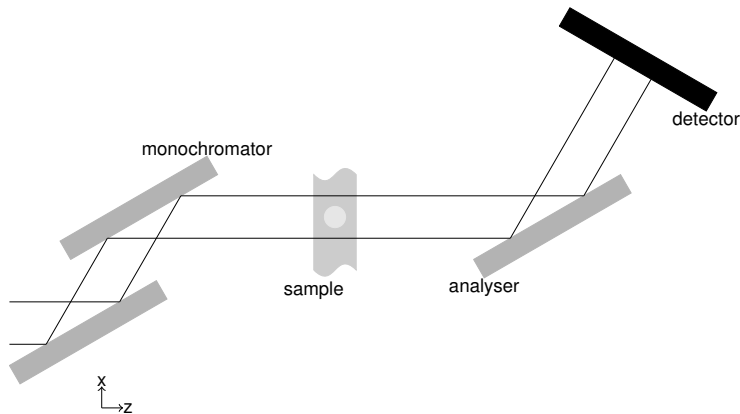


Figure 6: Example of analyser based imaging set-up.

195 modulated by changing the angle of incidence of the X-ray beam on the second  
 196 (analyser) crystal (see Figure 7a). This characteristic curve takes the name of  
 197 rocking curve and it is key in the image formation process. It results from the  
 198 combination of the reflectivity curves of both the monochromator and the anal-  
 199 yser crystal with a contribution arising from the beam divergence [65]. When  
 200 the system is tuned in such a way that roughly half of the intensity reaches the  
 201 detector (at full width half maximum of the rocking curve), small changes in  
 202 the direction of the propagation of the X-rays due to refraction in the sample  
 203 are transformed into intensity changes at the image receptor. The change in  
 204 direction of propagation is directly proportional to the gradient of the sample's  
 205 phase [13]

$$\theta_R = \frac{\lambda}{2\pi} \frac{\partial \phi(x, y)}{\partial x} \quad (14)$$

206 and an imaging system where contrast is proportional to the refraction angle  
 207 is often referred to as differential phase-contrast imaging system. The image  
 208 recorded in the case of the sphere sample is shown in Figure 7b. The X-rays  
 209 going through the centre of the sphere experience little or no refraction at all,  
 210 therefore their direction of propagation is not changed and they are transmitted  
 211 by the analyser with the same probability of the radiation that is not hitting  
 212 the sample. The image contrast in this region of the sample is mainly due to

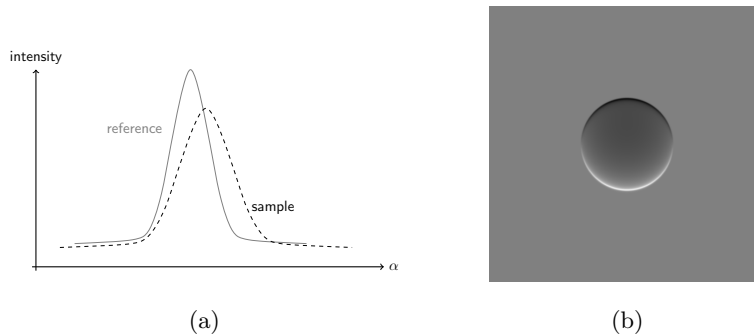


Figure 7: Image formation principle in an analyser-based system. (a) typical rocking curve: intensity recorder at a fixed position in the detector plane as a function of the “rocking” angle  $\alpha$  of the analyser crystal. (b) intensity recorded at the detector when the analyser crystal is set at fixed angle, in such a way that 50% of the intensity is transmitted (obtained by using the X-Tract software [12]).

213 X-ray absorption within the sample. Refraction increases while approaching the  
 214 sphere’s edges, where the change in the direction of propagation is maximum.  
 215 Because X-rays are deflected away from the beam axis (a glass sphere in air acts  
 216 as a diverging lens in the X-ray regime), the angle of incidence of the radiation on  
 217 the analyser will be changed in two opposite ways at the two edges of the sphere.  
 218 On one side, this will result in a higher probability of transmission through the  
 219 analyser, while it will translate into a smaller transmission probability on the  
 220 opposite side. This is the mechanisms at the basis of the generation of the dark  
 221 and bright fringes of Figure 7b.

222 Early implementations of this technique for imaging fusion pellets are those  
 223 of Goetz and Forster [66, 67]. This approach became increasingly popular after  
 224 1995 [68, 69] when methods to quantitatively separate phase and absorption  
 225 contributions were developed [70, 71]. This method is intrinsically sensitive to  
 226 the phase gradient in a single direction only and an additional measurement  
 227 is tipycally required to quantify the other component [72]. Another key de-  
 228 velopment that soon followed was the possibility to quantify the effect of the  
 229 scattering in the sample on the width of the rocking curve, which was put in  
 230 relation to sub-pixel scale features [73–77]. This was also extended for applica-

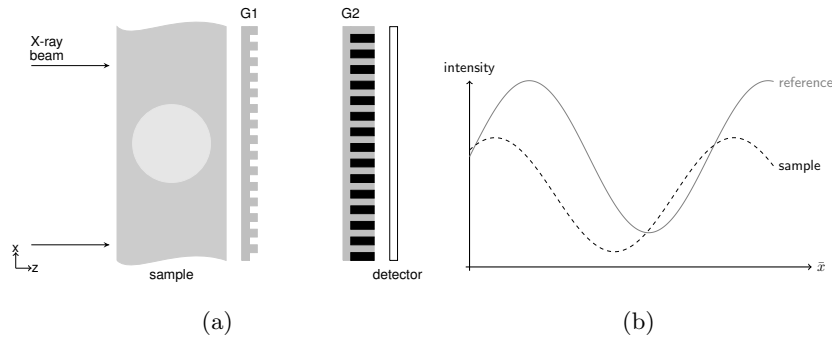


Figure 8: Example of grating based imaging set-up. (a) typical arrangement where the phase grating is placed after the sample and the (analyser) amplitude grating immediately precedes the detector. (b) intensity recorded at a fixed  $(x, y)$  position in the detector plane as a function of the scanning position of one grating relative to the other, along the  $x$  direction.

231 tions to tomography [78]. Methods for retrieving phase information from the  
 232 simultaneous acquisition of two images have been developed for the case of Laue  
 233 analyser [79].

234 Some examples of the many applications of analyser based techniques in  
 235 the medical field [80] are: cartilage [81–83], musculoskeletal [84] and breast  
 236 tissue [65, 85–87] imaging and dynamic tracking of micro-bubble concentrations  
 237 [88, 89].

### 238 3.4. Grating based imaging

239 Grating-based imaging methods make use of periodic structures to condition  
 240 and analyse the X-ray beam. A typical embodiment of this technique is sketched  
 241 in Figure 8a where the X-ray beam traverses the sample that modulates its  
 242 amplitude and imposes phase shifts. It is then passed through the phase grating  
 243  $G_1$  and analysed by the absorption grating  $G_2$  immediately before the image  
 244 receptor. If one of the two grating is laterally scanned (along  $x$ ) without the  
 245 sample in the beam, a modulated intensity curve is detected in each pixel. This is  
 246 often referred to as the phase-stepping curve. When the sample is present in the  
 247 beam, this curve is modified in three ways, as depicted in Figure 8b. The relative  
 248 reduction of the baseline is the conventional absorption image, the lateral shift of

249 the curve represents the differential phase contrast and the reduction of visibility  
 250 is linked to the scattering in the sample, or dark-field imaging. This can be  
 251 expressed quantitatively by writing the intensity oscillations recorded at a point  
 252  $(x, y)$

$$I(\bar{x}; x, y) = \sum_i a_i(x, y) \cos\left(\frac{2\pi i \bar{x}}{p_2} + \Phi_i(x, y)\right) \quad (15)$$

$$\approx a_0(x, y) + a_1(x, y) \cos\left(\frac{2\pi i \bar{x}}{p_2} + \Phi_1(x, y)\right) \quad (16)$$

253 where  $p_2$  is the period of  $G_2$  and  $a_i$  and  $\Phi_i$  are the amplitude and phase coefficient,  
 254 respectively. The images of the sample are reconstructed by comparing  
 255 the phase-stepping curves recorded without  $^w$  and with  $^o$  the sample in the  
 256 beam [90]. The transmission image, analogous to the one obtained in conventional  
 257 radiography, is given by

$$T(x, y) = a_0^o(x, y)/a_0^w(x, y). \quad (17)$$

258 The differential phase-contrast projection image of the sample is calculated by  
 259 taking the difference  $\nabla_x \phi(x, y) = \nabla_x^o \phi(x, y) - \nabla_x^w \phi(x, y)$ , and by considering  
 260 that

$$\nabla_x \phi(x, y) = \frac{p_2}{\lambda d} \Phi_1(x, y) \quad (18)$$

261 where  $d$  is the distance between  $G_1$  and  $G_2$ . Dark-field images are obtained by  
 262 first computing the normalized oscillation amplitude

$$V^w(x, y) = \frac{I_{max}^w(x, y) - I_{min}^w(x, y)}{I_{max}^w(x, y) + I_{min}^w(x, y)} \quad (19)$$

$$= \frac{a_1^w(x, y)}{a_0^w(x, y)} \quad (20)$$

263 and then by taking the ratio of this quantity, with and without sample in the  
 264 beam

$$V(x, y) = \frac{V^o(x, y)}{V^w(x, y)} \quad (21)$$

$$= \frac{a_1^o(x, y)a_0^w(x, y)}{a_1^w(x, y)a_0^o(x, y)} \quad (22)$$

265 which does not show changes ( $V(x, y) = 1$ ) for samples with negligible or absent  
266 small-angle scattering and is reduced ( $V(x, y) < 1$ ) when scattering occurs.

267 Introduction of grating-based techniques could be dated back to early '90s  
268 [91, 92] with experiments following few years later [93–97]. A breakthrough  
269 for the diffusion of this technique was the introduction of a third grating that  
270 enabled the use of low brilliance sources [98], tomography [99] and dark-field  
271 [100] or scattering imaging were also developed soon afterwards with a three-  
272 grating setup. An alternative method was subsequently proposed for differential  
273 phase-contrast imaging with weakly coherent hard X-rays [101]. Quantitative  
274 three-dimensional dark-field imaging was then developed for these grating-based  
275 imaging set-ups [102, 103]. The simultaneous determination of the two compo-  
276 nents of the phase gradient, by means of gratings structured in two dimen-  
277 sions, has also been discussed [104–107]. Another important development is  
278 the inverse geometry [108] which can enable compact Talbot-Lau interferome-  
279 try setups [109]. A much more detailed review of advances and milestones of  
280 grating-based X-ray phase-contrast imaging can be found in a recent review [90].  
281 A recent and very promising development on this front was the successful fab-  
282 rication of grating structures with approximately one order of magnitude finer  
283 pitch [110] and their application for the enhancement of table-top imaging sys-  
284 tems [111]. Albeit based on gratings, the working principle of this approach is  
285 different from that of the more conventional grating-based interferometry and  
286 it is best understood under the concept of univesal moiré effect [112]. This  
287 eliminates the need for an absorption grating and enables the realization of  
288 a polychromatic far-field interferometer that can overcome the limitations in  
289 sensitivity and dose efficiency of more conventional bench-top interferometers  
290 [112].

291 An extremely wide spectrum of application exists also for grating based  
292 imaging techniques, examples are: breast tissue [113–115], brain tumour [116],  
293 cartilage [117–120], and lungs [121, 122]. A fairly recent review exists that is  
294 fully dedicated to this topic [123].

295 *3.5. Tracking based methods*

296 Another broad category of X-ray phase-contrast imaging techniques stems  
297 from the observation that it is possible to measure the sample absorption, re-  
298 fraction and scattering by imposing a known structure to the radiation field and  
299 by directly tracking its modifications. In general terms, an overall reduction of  
300 the structured beam intensity can be traced back to absorption of the radiation  
301 within the sample, while the spatial distortions of the known intensity patterns  
302 are used to infer the phase shifts imposed by the sample to the wavefront.

303 The introduction of this approach may be traced back to the ‘90s [124, 125]  
304 with experiments following several years later. The structuring could be imposed  
305 by using a lenslet array [126], a microprobe [127], an absorption grid [128–  
306 130], a phase grating [131] or a speckle pattern [132, 133] and the distortions  
307 imposed by the sample can be tracked by using a high-resolution detector, also  
308 in combination with sub-pixel resolution analysis [134] or by using Fourier-based  
309 analysis [135]. These approaches can be extended to two-dimensional sensitivity  
310 [136–138], to include dark-field contrast [139, 140] and directional dark-field  
311 imaging [141–143]; as well as three-dimensional imaging with tomography [142,  
312 144].

313 Applications of tracking based techniques include: bone imaging [145], dy-  
314 namic airways imaging [146, 147] and metrology [142, 148, 149].

315 **4. Edge illumination**

316 Edge illumination X-ray phase-contrast imaging has been investigated in  
317 the recent years as a possible way forward for the translation of phase-sensitive  
318 imaging techniques into mainstream applications. Edge illumination was ini-  
319 tially developed in synchrotron experiments at Elettra (Italy) and was inspired  
320 by analyser-based methods [150]. The typical experimental set-up is reported in  
321 Figure 9a. A beam of synchrotron radiation, propagating from left to right, is  
322 shaped down to a narrow blade of radiation by an aperture. It then traverses the  
323 sample and impinges on the edge of a second aperture that is placed in front of



324 the image receptor. If one of the two apertures is laterally shifted (along  $x$ ) the  
 325 recorded intensity is modulated: it reaches a maximum when the two apertures  
 326 are perfectly aligned and it progressively decreases for increasing lateral shifts  
 327 (see Figure 9b). This is often called illumination function and characterises the  
 properties of this type of imaging systems.

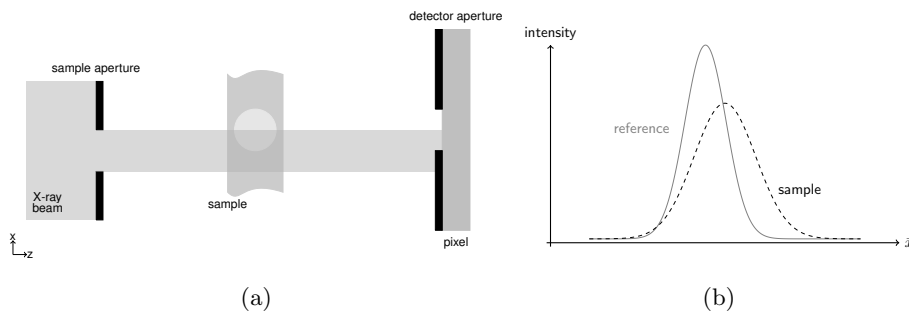


Figure 9: Edge illumination working principle: (a) typical synchrotron set-up and (b) illumination function.

328  
 329 The working principle of edge illumination can be explained by observing  
 330 that refraction in the sample results in lateral shifts of the X-ray beam which  
 331 are translated into intensity modulations by the presence of the second aper-  
 332 ture. Referring to Figure 9a, a deflection upwards will result in an increased  
 333 intensity at the detector pixel while a decreased intensity would be recorded if  
 334 the deflection occurs downwards. This holds for a completely transparent object  
 335 that only perturbs the phase of the X-ray beam. If the sample is also absorbing,  
 336 then at least two images have to be acquired to extract the sample's absorption  
 337 and refraction [151]. This is typically achieved by recording two intensity pro-  
 338 jections, with the apertures aligned in such a way that the shaped X-ray beam  
 339 impinges on the two edges of the detector aperture. If the apertures are aligned  
 340 such that half of the intensity reaches the detector in both cases, these two  
 341 configurations correspond to the two points at the full width half maximum of  
 342 the illumination function (see Figure 9b). The edge illumination principle can  
 343 also be implemented with a laboratory set-up that uses rotating anode X-ray  
 344 tubes with extended focal spots [152] (sketched in Figure 10). The diverging and

345 polychromatic beam generated by this type of sources is shaped by a pre-sample  
 346 mask that creates a series of independent beamlets. These propagate through

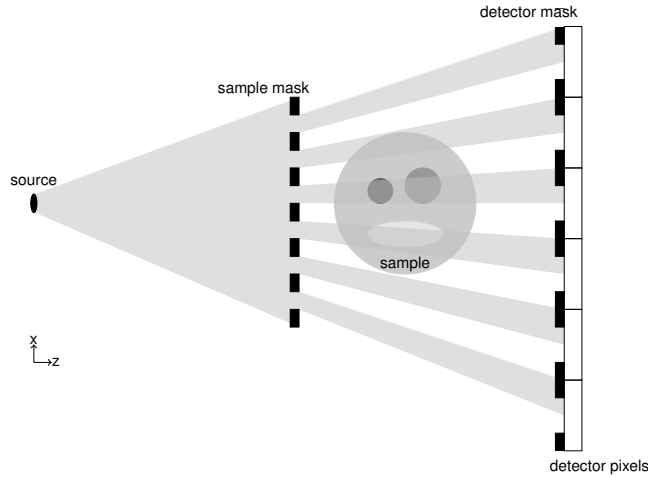


Figure 10: Laboratory set-up for edge illumination X-ray phase-contrast imaging.

346

347 the sample and are then analysed by a second set of apertures before the de-  
 348 tector. The pitches of both the pre-sample and the detector mask are harmoni-  
 349 cally matched to that of the detector pixels such that a one-to-one relationships  
 350 exists between each aperture in both masks and each detector pixel column  
 351 (along  $y$ ). This approach has negligible spatial or temporal coherence require-  
 352 ments [153, 154], provides high sensitivity also for laboratory implementations  
 353 [155, 156], enables the simultaneous attainment of high sensitivity and dynamic  
 354 range [157], is robust against thermal and mechanical instabilities [158, 159] and  
 355 the set-up can be made compact [160, 161]. By using a microfocal source it is  
 356 possible to adopt a large magnification geometry and perform hard X-ray phase  
 357 imaging with micrometre resolution [162]. Two-dimensional sensitivity can be  
 358 simultaneously achieved by using masks structured in two dimensions [163].

359 Dark-field images can be quantitatively retrieved by acquiring (at least) a  
 360 third intensity projection [164, 165] and by using a Gaussian representation  
 361 of the intensity. Under general conditions, the illumination function  $L(\vec{x})$  (see

362 Figure 9b) can be expressed in the following way

$$I(\bar{x}) = \sum_m \sum_n A_{mn} \exp \left[ -\frac{(\bar{x} - \mu_{mn})^2}{2\sigma_{mn}^2} \right] \quad (23)$$

363 where  $\mu_{mn} = \mu_m + \mu_n$ ,  $\sigma_{mn}^2 = \sigma_m^2 + \sigma_n^2$  and  $A_{mn} = A_m A_n (1/\sqrt{2\pi\sigma_{mn}^2})$ . Both  
 364 the illumination function  $L(\bar{x}) = \sum_n (A_n/\sqrt{2\pi\sigma_n^2}) \exp[-(\bar{x} - \mu_n)^2/2\sigma_n^2]$  and  
 365 the object function  $O(\bar{x}) = \sum_m (A_m/\sqrt{2\pi\sigma_m^2}) \exp[-(\bar{x} - \mu_m)^2/2\sigma_m^2]$  have been  
 366 represented as the sum of Gaussian functions, ( $m = 1 \dots M$  and  $n = 1 \dots N$ ).  
 367 A single-Gaussian representation of both illumination and object function is  
 368 accurate in many practical cases and this allows for an analytic solution of  
 369 Equation 23 [164]. Should this not be the case, the number of terms to be  
 370 retained in Equation 23 can be increased and the sample's parameters retrieved  
 371 numerically [159, 166].

372 Tomographic edge-illumination X-ray phase-contrast imaging was developed  
 373 at synchrotron sources [167] and adapted to rotating anode tubes [168, 169],  
 374 including three-dimensional dark-field imaging [157]. A reverse-projection re-  
 375 construction method enabled a step change in the data acquisition strategy  
 376 by allowing continuous rotation of the sample [170, 171]. More recent develop-  
 377 ments include algorithms for robust reconstructions [172, 173] and a single-image  
 378 phase retrieval algorithm [174] that, albeit requiring homogeneity of the sam-  
 379 ple, greatly simplifies the practical implementation of the method especially with  
 380 respect to tomography [175]. This can be extended to include multi-material  
 381 samples [176].

382 Examples of use in applied investigations of edge illumination X-ray phase  
 383 contrast imaging are: low-dose mammography [177, 178], cartilage imaging [179,  
 384 180], security [181], baggage screening [182] with a large field of view scanning  
 385 system [183–185], composites materials [186, 187], regenerative medicine [188]  
 386 and lung imaging [189].

387 **5. Conclusion**

388 X-ray phase-contrast imaging can extend the applicability of radiography  
389 and tomography for visualising the internal structure of samples that do not  
390 exhibit enough absorption contrast. Various methods have been developed to  
391 obtain phase contrast images in the hard X-ray regime, and they were intro-  
392 duced and described along with examples of applications. The edge illumination  
393 approach, that has been subject of investigation and developments by our group  
394 in the recent years, was finally presented and discussed.

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