A single-image method for x-ray refractive index CT

To cite this article: A Mittone et al 2015 Phys. Med. Biol. 60 3433

View the article online for updates and enhancements.
A single-image method for x-ray refractive index CT

A Mittone\textsuperscript{1,2,4}, S Gasilov\textsuperscript{1,3}, E Brun\textsuperscript{1,4}, A Bravin\textsuperscript{4} and P Coan\textsuperscript{1,2}

\textsuperscript{1} Department of Physics, Ludwig Maximilian University, 85748 Garching, Germany
\textsuperscript{2} Department of Clinical Radiology, Ludwig Maximilian University, Munich 81377, Germany
\textsuperscript{3} ANKA Synchrotron Radiation Facility, Karlsruhe Institute for Technology, Eggenstein 76344, Germany
\textsuperscript{4} European Synchrotron Radiation Facility (ESRF), Grenoble 38043, France

E-mail: alberto.mittone@esrf.fr

Received 15 December 2014, revised 5 February 2015
Accepted for publication 12 February 2015
Published 9 April 2015

Abstract

X-ray refraction-based computer tomography imaging is a well-established method for nondestructive investigations of various objects. In order to perform the 3D reconstruction of the index of refraction, two or more raw computed tomography phase-contrast images are usually acquired and combined to retrieve the refraction map (i.e. differential phase) signal within the sample. We suggest an approximate method to extract the refraction signal, which uses a single raw phase-contrast image. This method, here applied to analyzer-based phase-contrast imaging, is employed to retrieve the index of refraction map of a biological sample. The achieved accuracy in distinguishing the different tissues is comparable with the non-approximated approach. The suggested procedure can be used for precise refraction computer tomography with the advantage of a reduction of at least a factor of two of both the acquisition time and the dose delivered to the sample with respect to any of the other algorithms in the literature.

Keywords: x-ray imaging, phase-contrast, tomographic imaging

(Some figures may appear in colour only in the online journal)

1. Introduction

A very powerful method to study the internal structure and composition of an object is provided by x-ray computed tomography (CT). Currently, x-ray CT is widely employed in many different fields including material science, medical diagnostics, archeology and many others. In conventional CT the image contrast is determined by variations of the linear attenuation
coefficient within the object. A different approach is offered by the phase-contrast imaging (PCI) techniques. PCI allows us to quantify the refraction of x-rays in the sample and to obtain images whose contrast originates from the fluctuations of the refractive index inside the sample. PCI may offer many advantages with respect to standard absorption-based imaging (Bravin et al. 2013); in particular, the quantitative 3D CT reconstruction of the index of refraction (indicated in this work as RCT) can provide superior quantitative results in terms of both accuracy and image quality (Gasilov et al. 2013). The first step for performing RCT is to separate the refraction signal (i.e. to calculate the ‘differential phase’ image) from the other contrast contributions (i.e. absorption, ultra-small angle scattering [USAXS]) (Rigon et al. 2003), which are present in a raw phase-contrast image. This can be obtained by using different PCI techniques and algorithms (Bravin et al. 2013). Among others, the diffraction-enhanced imaging method (Chapman et al. 1997) and its extended version EDEI, (Rigon et al. 2003, Maksimenko 2007) can be applied in combination with the analyzer-based imaging (ABI) technique (Diemoz et al. 2010), the phase-stepping procedure is employed for grating interferometry (Weitkamp et al. 2005), the method proposed by Cloetens (Cloetens 1999) for the propagation-based technique and other approaches are used with the coded aperture imaging technique (Munro et al. 2012). However, all these methods require the acquisition and processing of multi- (at least two) image data sets, with a consequent increase of the delivered dose and of the required scan time. Efforts have been made to reduce the number of images needed down to one, when some additional information or assumption on the sample composition is known or boundary conditions are set (Paganin et al. 2002, Diemoz et al. 2011). We will focus here on the ABI, theoretically the most sensitive technique (Diemoz et al. 2010), for which no single image approach has yet been established.

The main goal of this work is to perform accurate CT reconstruction of refractive index using one single set of images resulting in an evident benefit in terms of delivered dose and acquisition time.

We suggest approximating the sample absorption by using a simple analytical function, and thus reducing the number of raw phase-contrast images required for the retrieval of the differential phase image. We have combined this approximation with the EDEI algorithm (we will refer to the here-proposed method as AEDEI), which allowed us to use a single set of PCI CT data acquired at only one position of the ABI analyzer crystal (AC) to obtain differential phase images. It is shown that the approximate differential phase images extracted with the AEDEI method can be used for an accurate reconstruction of RCT images. We tested the suggested approximation by using a tomographic set of ABI images in a thick biological sample (i.e. a whole breast tissue). The extracted approximated differential phase images have then been used for the CT reconstruction of the index of refraction within the sample.

2. Mathematical description

The complex index of refraction defined as $n = 1 - \delta + i\beta$, where $\delta$ represents the index of refraction decrement and $\beta$ is the absorption component, can be used to describe the interaction of x-ray with matter. Under the geometrical optics approximation, several algorithms exist to separate the different contributions to the signal on the detector. Let us consider an ABI setup and approximate the rocking curve (RC) of the AC with a Gaussian distribution using the method proposed by Hu et al. (Hu et al. 2008). After x-rays have passed through the object, the intensity recorded by the detector $I_t$ can be expressed using equation (1), where $I_t$ represents the intensity transmitted after the sample, $\Delta \theta$ is the refraction angle and $\theta_{RC}$ is the angular displacement of the AC with respect to the Bragg angle for the selected x-ray energy.
In order to separate the absorption ($I_t$) and refraction components ($\Delta \theta$), several images are normally required. Among all the methods, the DEI and EDEI algorithms require the least number of raw phase-contrast images for extracting maps of absorption and refraction. Two images (or sets of images in CT) are acquired at symmetric angular positions of the AC with respect to the center of its RC. In order to perform accurate CT reconstruction of the index of refraction using one single set of images we assume that the object is homogenous and it produces an absorption signal $I_t$, which is solely determined by the object’s geometry. The x-ray intensity downstream from our object can be calculated pixel by pixel using the well-known Lambert–Beer law $I(x, y) = I_0(x, y) e^{-\mu t(x,y)}$, where $t(x,y)$ is the thickness at the point $(x,y)$ projected on the exit plane of the object, $\mu$ is the absorption coefficient and $I_0(x,y)$ is the reference incoming x-ray intensity.

The assumption of a homogeneous sample is verified for breast tissues: the differences in the absorption coefficients between the tissues composing the organs do not exceed a few percent (www.nist.org). The same assumption of a homogenous object can be applied in other organs, like the brain, kidney, etc.

Using this approximation for the object absorption, from equation (1) the distribution of the refraction angle can be calculated as:

$$\Delta \theta(x, y) = \sqrt{-2\sigma^2 \log \left( \frac{I(x, y)}{I_t(x, y)} \right)} - \theta_{RC}. \quad (2)$$

Equation (2) allows the calculation of the refraction angle projections starting from the projections acquired at one angular position of the AC along its RC. For the validity of equation (2) several experimental conditions have to be fulfilled: the radius of the first Fresnel zone has to be small compared to the characteristic size of the object’s inhomogeneities or to the detector pixel size (i.e. the diffraction inside the object can be neglected); the inspected object has to present small variations in the linear attenuation coefficient and the USAXS component which gives rise to a broadening of the rocking curve which can be neglected. Both conditions, already assumed by several other algorithms used to separate the different components of the signals (Paganin et al 2002, Maksimenko 2007, Diemoz et al 2011), are satisfied for many objects when imaging is performed using hard x-rays. The projections computed as in equation (2) can then be used to perform the CT reconstruction of the fluctuations of the index of refraction $\delta$. When valid, the suggested approximation for the sample absorption can be used to reduce the number of raw phase-contrast images, which should be combined to retrieve the differential phase, as well as in other cases and imaging techniques. Consequently, the dose delivered to the sample and the time of phase-contrast tomographic data acquisition can be significantly reduced, which are very important variables in imaging biological samples.

### 3. Experiment and results

The here-proposed approach has been tested on breast imaging data acquired in an ABI CT experiment at the ID17 beamline of the ESRF. A double Si(111) Laue crystal was used to select a quasi-monochromatic ($\Delta E / E \sim 10^{-4}$) and quasi-parallel x-ray wave (horizontal divergence $<1$ mrad, vertical $<0.1$ mrad) from the beam produced by a 21-pole wiggler. Images were acquired using a monochromatic beam of 51 keV. The x-ray beam after propagation through the sample was analyzed by a 3 cm thick, symmetrically cut Si(333) crystal. A second identical crystal placed upstream from the sample was used as a post-monochromator (figure 1). The experimental RC has been measured and its full width at half maximum was...
estimated to be 1.7 µrad. The best possible contrast is achieved where the slopes of the AC’s RC are maximized. Two CT data sets at two symmetrical positions along the AC’s RC, were acquired at θ_{RC,1,2} = ±0.85 µrad with respect to the Bragg angle (i.e. center of the RC). Only one set of images has been used for the AEDEI method, while both sets have been considered for the application of the standard EDEI. The EDEI results have been used for comparison and for verifying the accuracy of our approximated approach. To reconstruct the index of refraction of CT images we choose here the filtered back-projections for gradient projections applied with the Hilbert filter (Faris and Byer 1988) but other algorithms can be used as well (Gasilov et al 2014). The number of angular views (projections) uniformly acquired in a range of angles (0, π] was 500. However, similar results are obtained using 250 projections when iterative CT reconstruction algorithms are employed (Zhao et al 2012). The detection system used to record the images was a charge-coupled detector (CCD) with a gadolinium-based fluorescent screen having an effective pixel of 96 × 96 µm² (FReLoN detector, bin 2 × 2 (Coan et al 2006)).

The investigated sample consisted of an excised human breast tissue, formalin-fixed, of 7 cm in diameter. It was acquired through the Institute of Gynaecology of the Ludwig-Maximilians-University, Munich. The full study was conducted in accordance with the Declaration of Helsinki and was approved by the local ethics committee. The patient’s written informed consent was obtained before inclusion after adequate explanation of the study protocol.

The sample was positioned in a cylindrical PMMA container with a wall thickness of 0.5 cm. The data acquisition was performed at the biomedical beamline (ID17) of the European Synchrotron Radiation Facility (ESRF, Grenoble, France). In order to analytically calculate the x-ray absorption, we approximated our sample with a homogenous cylinder made of breast tissue, of diameter equivalent to our real sample, and we used the absorption coefficient for breast provided by the NIST database (www.nist.org).

The reconstructed fluctuations of the index of refraction, \( \tilde{\delta} \), have been converted into density values by using the electrodynamics expression \( \rho \sim k\tilde{\delta} \), where \( k \) is a material-independent constant at a given energy (Landau and Lifshitz 1984).

Figures 2(a)–(c) report the comparison of the CT slices reconstruction using: (a) the refraction angle projections obtained with the EDEI algorithm (i.e. using two data sets at two angular positions on the RC), (b) the projections obtained with the method here proposed (AEDEI), and (c) the experimental projections without extraction of the refraction signal (i.e. mixed signals). Two magnified regions of the axial views are presented for the EDEI and AEDEI cases (figures 2(d)–(g)), respectively.
The mixed signal images have been reconstructed by applying the filtered back-projection algorithm, with the ramp filter, directly on the raw phase-contrast images. It has been shown that the CT reconstruction of the mixed signal (absorption and refraction) performed using one single set of images can already provide useful clinical information owing to the edge enhancement introduced by using the ABI technique (Sztrókay et al 2011). However, this approach in certain cases is not sufficient for a clear discrimination of some details in the investigated object (figure 2(c)). A superior discrimination of these fine details can instead be achieved by using the CT reconstruction together with the AEDEI approach using the same set of images (figure 2(b)).

The comparison of the CT reconstruction results in the sagittal direction for the EDEI and AEDEI is presented in figure 3. The final reconstructed images obtained with the AEDEI have been corrected because they are affected by a linear gradient along the horizontal direction. This systematic error appears because of a non-perfect normalization of the ABI projections.
with respect to the normalized RC. The correction has been performed by calculating the
derivative of the gradient and by applying a corrective mask. The mask linearly equalizes the
values obtained at the two extremes of the image (left and right border). The obtained density
values are then normalized on a reference value (taken within the wall of the PMMA container
in our case).

Negligible qualitative differences in distinguishing fine details have been detected in both
the axial (figure 2) and the sagittal views (figure 3). Besides, the AEDEI-reconstructed CT
images present a signal-to-noise ratio, calculated in the region marked by the red crosses in
figure 2(a), about 40% higher than the EDEI-reconstructed ones. In contrast, a decrease of
the image contrast between the different tissues of about a factor of 3 can be measured when
the AEDEI approach is used figures 3(c) and (d). The increase of the signal-to-noise ratio and
the reduction of the contrast are related to the smoothing of both the signal and noise induced
by the approximation introduced in the AEDEI. From a quantitative point of view, as visible
from the profiles of figures 3(c) and (d), the signal obtained using the AEDEI presents smaller
fluctuations in contrast to the one obtained with the standard EDEI. For a given tissue, the den-
sity values obtained with the AEDEI differs up to 15% with respect to the values calculated
with the EDEI. This fact is strictly related to the normalization of the experimental projection
images with respect to the corresponding intensity of the RC of the AC.

From a qualitative point of view, the results obtained from AEDEI-CT reconstructions are
comparable to the EDEI-CT ones. In both cases it is possible to easily distinguish the different
types of tissues like the adipose tissue (darker regions), the glandular strands (brighter internal
regions) and the skin (external white region). The black areas visible inside the breast are air
bubbles.

The gradient visible in the profile of figure 3(d), AEDEI case (black line), is related to the
non-perfect correction by the mask mentioned above. This approach has been applied to dif-
ferent breast samples, presenting results of the same quality as those here reported.
4. Conclusions

In summary, the proposed AEDEI method is useful for qualitative investigations because the image quality is preserved despite the approximated approach. The remarkable characteristic of AEDEI is that it requires only a single image to retrieve the phase with respect to the two or more requested by the other methods reported in the literature. The present limitations of AEDEI are in the quantitative significance of data, due to the approximation of the absorption and the flat-field normalization of the experimental images. Moreover, the presence of image ‘artifacts’, like the linear gradient appearing in the CT reconstructions, may affect the quantitative results. It is worth mentioning that the same principle on which this approach relies can be applied for phase retrieval using other PCI techniques after having adapted the algorithms used to separate the different contributions to the signal recorded by the detector.

This single image method may be of great interest in studying biological tissues because of the reduction of the delivered dose and of the acquisition time. Also, it is probably a unique method applicable to living samples, where the perfect repositioning of multiple images is unfeasible.

Acknowledgments

The authors would like to acknowledge financial support from the Deutsche Forschungsgemeinschaft cluster of excellence, the Munich Center for Advanced Photonics (EXE158) and the ESRF for the provision of beamtime; and H Requardt, T Brochard, C Nemoz and the rest of the ID17 staff for the technical support at the beamline.

References

Hu C, Zhang L, Li H and Lo S 2008 Comparison of refraction information extraction methods in diffraction enhanced imaging Opt. Express 16 16704
Landau L D and Lifshitz E M 1984 Electrodynamics of Continuous Media (New York: Pergamon)
Maksimenko A 2007 Nonlinear extension of the x-ray diffraction enhanced imaging Appl. Phys. Lett. 90 154106