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A software platform for phase contrast x-ray breast imaging research

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ABSTRACT

Purpose: To present and validate a computer-based simulation platform dedicated for phase contrast x-ray breast imaging research.
Methods: The software platform, developed at the Technical University of Varna on the basis of a previously validated x-ray imaging software simulator, comprises modules for object creation and for x-ray image formation. These modules were updated to take into account the refractive index for phase contrast imaging as well as implementation of the Fresnel–Kirchhoff diffraction theory of the propagating x-ray waves. Projection images are generated in an in-line acquisition geometry. To test and validate the platform, several phantoms differing in their complexity were constructed and imaged at 25 keV and 60 keV at the beamline ID17 of the European Synchrotron Radiation Facility. The software platform was used to design computational phantoms that mimic those used in the experimental study and to generate x-ray images in absorption and phase contrast modes.
Results: The visual and quantitative results of the validation process showed an overall good correlation between simulated and experimental images and show the potential of this platform for research in phase contrast x-ray imaging of the breast. The application of the platform is demonstrated in a feasibility study for phase contrast images of complex inhomogeneous and anthropomorphic breast phantoms, compared to x-ray images generated in absorption mode.
Conclusions: The improved visibility of mammographic structures suggests further investigation and optimisation of phase contrast x-ray breast imaging, especially when abnormalities are present. The software platform can be exploited also for educational purposes.

1. Introduction

X-ray breast imaging modalities like the traditional two-dimensional (2D) mammography and the advanced three-dimensional (3D) breast tomosynthesis use low-dose projection images of the breast acquired with dedicated x-ray based mammography systems. In 2D mammography, the typical breast screening exam includes two x-ray images of each breast taken at craniocaudal and mediolateral oblique views. In breast tomosynthesis, a limited number of x-ray projection images of the breast (typically 9 to 25) are acquired, while the x-ray source and the detector rotate along an arc around the breast. The projection images are then used with reconstruction techniques to provide a pseudo 3D image of the breast. In such exams, the total absorbed dose in the breast is typically in the order of a few mGy (0.6 to 4.0 mGy) [1,2]. Each mammogram presents a 2D distribution of the intensity of the x-rays reaching the detector. This type of projection imaging is based on the absorption properties of the various breast tissues and when applied to dense breasts, it may impair the task of diagnosing tumour-like structures due to the superposition of large areas of normal glandular tissue. The overlapping of tissue structures is partly removed in the 3D tomosynthesis acquisition. However, an additional limitation of such 2D or 3D imaging modalities applied to the breast, is that they are based on x-ray absorption in the breast, where attenuation coefficients of normal and cancerous breast structures (like invasive/non-invasive ductal and lobular carcinoma) are very close and the resulting subject contrast is less than 10%, typically [3,4].

Phase-contrast (PhC) imaging uses a different approach that has proved to increase the visibility of internal details in objects (e.g. biological tissues) composed of elements with low atomic number [5]. PhC exploits the differences between the refractive index of different materials to improve the visibility between these structures in 2D x-ray images. Different techniques have been developed for detecting PhC effects. In the in-line (or so-called propagation based) PhC imaging, contrast enhancement effects are

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produced with an x-ray source characterized by a certain degree of lateral spatial coherence, by recording the diffracted and refracted wave field downstream of the irradiated object, with a suitable high resolution area detector [5,6]. In such a scheme, the visual appearance of PhC effects in the final image is a characteristic contrast enhancement at the interfaces between structures with different x-ray refractive indices.

The clinical implementation of every new imaging modality, however, is preceded by an optimisation work focussed on optimal energy and acquisition geometries which both depend on the sample, type of the detectors and phase retrieval algorithms in respect to the visibility of structures of interest (breast lesions). The best approach to carry out optimisation work is by the use of computer modelling and simulation.

Most computer simulations performed in the field of PhC are limited to the use of simple phantoms, composed of few geometrical forms like cylinders and spheres that approximate fibers and tumours in the breast tissue. These geometries are adopted in investigations on the optimal geometrical distances like the distance object-to-detector as a function of the object shape and its size [7–11]. Investigations also focussed on the influence of the pixel dimensions on the PhC signal [10], on the development of phase retrieval algorithms [11], to study the suitability of inverse Compton sources for PhC mammography [12], on comparison of several methods for phase retrieval for CT imaging [13,14]. All these are carried for in-line phase contrast. These simple phantoms play an important role in setting and optimising the imaging technique. Computational phantoms characterised with increased complexity are also required to test and evaluate the performance of advanced 3D imaging techniques. In the case of breast imaging, the use of advanced breast phantoms would allow to perform accurate breast dosimetry, to investigate the effect of different imaging acquisition parameters on the quality of breast structures on images, to test and optimise novel and existing reconstruction techniques, etc. Recent simulations in the field of in-line PhC mammography were performed with an average compressed breast modelled as a homogeneous slab and abnormalities represented by spheres and cylinders with different diameters [12,15–17]. Further investigations in this field, however, require simulations with more complex, anthropomorphic breast phantoms.

X-ray PhC effects can be simulated in several ways. One possible approach is the analytical one, based on the wave-optical theory (Fresnel–Kirchhoff diffraction integral and paraxial approximation) [18]. This method allows evaluating the diffraction angle of photons exiting from the object by means of simple equations [19]. In some papers, the Wigner distribution formalism was applied to modelling the phase effects [20]. Another approach is the use of an advanced wave-optical method proposed by Bravin et al. [21]. The approach was applied successfully to simulate diffraction enhanced x-ray imaging with large and thick objects, eliminating the assumption of the usual 'weak object approximation'. Monte Carlo methods were utilised to simulate the refractive x-ray interactions [15,22,23]. However, the use of Monte Carlo techniques is still a challenge in the field of PhC imaging and simulations are still at the initial stage.

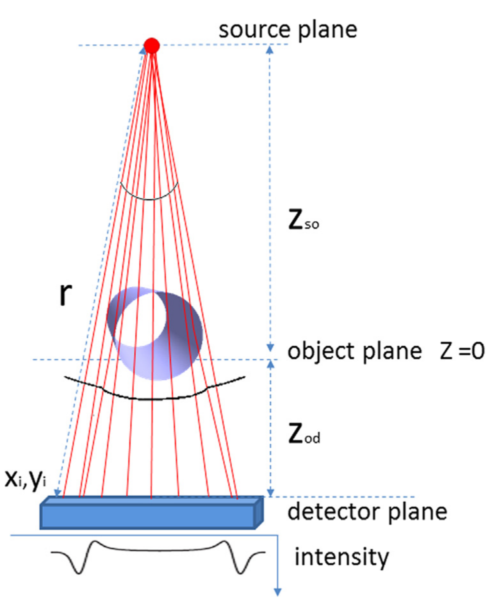

Fig. 1. Schematic representation of the mechanism of edge enhancement due to refraction of x-rays at the edges of an object in the in-line propagation, used as a model to generate PhC images.

There are few published simulation platforms for PhC dedicated to 2D and 3D x-ray breast imaging. The most complete one is the Virtual X-ray Imaging (VXI) [9,19], based on a computer code dedicated to the simulation of radiographic, radioscopic or tomographic procedures [24,25]. The PhC effects are simulated based on the Fresnel–Kirchhoff integral and they can be integrated with computer-aided design objects and objects based on non-uniform rational B-spline. The usefulness and application of this code have been shown by simulating simple objects like cylinders and spheres as well as objects of different shapes embedded into a 1.5-mm thick bounding box.

Our goal is to develop and evaluate algorithms for volumetric breast reconstructions from PhC x-ray projection images, obtained in a tomosynthesis setup. The first stage of this project aims at the development of a software tool for generation of PhC x-ray images from simple phantoms as well as from more complex breast-like models. This paper presents the results from the development, validation and the initial testing of the simulation platform dedicated to PhC breast imaging studies. The validation study involved comparison of simulated images of simple phantoms with those obtained from experimental work carried out at the European Synchrotron Radiation Facility (ESRF), Grenoble, France. The in-line (or so-called propagation based) acquisition geometry was used [6]. The developed platform was exploited to generate PhC x-ray images from complex inhomogeneous and anthropomorphic breast phantoms.

2. Materials and methods

2.1. The method used to generate PhC images

Fig. 1 shows the in-line configuration adapted for simulation of PhC x-ray images. The wave propagation is modelled based on the Fresnel–Kirchhoff diffraction theory [18]. A pure coherent radiation of wavelength λ is considered, emerging from a point-like monochromatic source, located at a distance z_{so} from the object plane. The object is placed in a plane located at $z=0$ and the detector is located at a distance z_{od} from the object plane.

The electric wave field in the free space, detected at a distance r from the source, has the following expression:

$$E(x_i, y_i) = \frac{A}{r} e^{ikr} = \frac{A}{r} e^{ik\sqrt{(x_i - x_s)^2 + (y_i - y_s)^2 + z^2}} \tag{1}$$

where A is the amplitude of the wave that originates in the source point (x_s, y_s) located in the source plane, (x_i, y_i) are the coordinates of the detection point on the detector plane, $r^2 = (x_i^2 + y_i^2)$, $z = z_{so} + z_{od}$, while $k = \frac{2\pi}{\lambda}$ is the wavenumber. In mammography, incident x-ray photons are in the energy range 10–32 keV, with a mean photon energy in the range 17–22 keV. For this range, the corresponding wavelength, λ varies between 56 pm and 73 pm. Since the tissue features of diagnostic interest in the breast have much greater size compared to these wavelengths, then one can