Published in final edited form as:

Proc SPIE Int Soc Opt Eng. 2012 ; 8313: . doi:10.1117/12.911422.

Investigation of Source Grating Stepping for Differential Phase-contrast Cone Beam CT (DPC-CBCT) System

Weixing Cai¹, Yang Yu², Ruola Ning^{1,2}, Jiangkun Liu², and David Conover³

¹Department of Imaging Sciences, University of Rochester Medical Center, Rochester, NY 14642

²Department of Electrical and Computer Engineering, University of Rochester, Rochester, New York, 14627

³Koning Corporation, 150 Lucius Gordon Drive, West Henrietta, New York 14586

Abstract

Differential phase contrast (DPC) imaging, which utilizes phase shift information of X-ray, has the potential of dramatically increasing the contrast in biological sample imaging compared to attenuation-based method that relies on X-ray absorption information, since the X-ray phase is much more sensitive than the attenuation during transmission. In a DPC imaging system, the phase stepping method is widely used to obtain DPC images: at each angle the phase grating is shifted incrementally to produce a set of images and then the so obtained images are used to retrieve DPC image. However, DPC imaging requires a high mechanical precision to perform phase stepping, which is generally one order higher than the period of phase grating. Given that phase grating period is generally 2–4 μm , the requirement of mechanical accuracy and stability are very demanding ($<0.5\mu\text{m}$) and difficult to meet in a system with rotating gantry. In this paper, we present a method that is able to greatly relax the requirement of mechanical accuracy and stability by stepping the source grating rather than the analyzer grating. This method is able to increase the system's mechanical tolerance without compromising image quality and make it feasible to install the system on a rotating gantry to perform differential phase-contrast cone beam CT (DPC-CBCT). It is also able to increase the grating shifting precision and as a result improve the reconstructed image quality. Mechanical tolerance investigation and image quality investigation at different phase stepping schemes and different dose levels will be carried out on both the original modality and the new modality, the results will be evaluated and compared. We will deliberately create random mechanical errors in phase stepping and evaluate the resulting DPC images and DPC-CBCT reconstructions. The contrast, noise level and sharpness will be evaluated to assess the influence of mechanical errors. By stepping the source grating, the system is expected to tolerate an error of 6–7 times bigger than that with analyzer grating stepping.

Keywords

Phase Contrast; Differential Phase Contrast; Phase Stepping; Cone Beam CT

1. INTRODUCTION

Differential phase contrast cone-beam CT (DPC-CBCT) is promising as the next breakthrough in the field of medical x-ray imaging and CT imaging [1–5]. Unlike

conventional CT, DPC-CBCT utilizes the phase information of x-rays, which is expected to be more sensitive than the attenuation in x-ray amplitude and to provide better details of contrast. It has the potential to further enhance both contrast resolution and spatial resolution in soft tissue imaging with a comparable dose level as that for absorption-based CBCT.

Phase stepping is widely used to retrieve the phase information, and literatures [1–5] suggest shifting either the phase grating or the analyzer grating N times ($N \geq 3$) to cover the distance of one period of the self image. At each step, an intensity image is acquired, and finally the N steps images are processed to retrieve the phase information. Due to the short coherence length of an x-ray system, the analyzer grating period normally resides in the proximity of several microns, and thus to shift the phase grating by precisely $1/N$ of the period requires high mechanical precision. In our system, the self image has a period of 4 micron, which means each step of phase shift is 0.5 micron for an eight-step scheme. Therefore the tolerance of mechanical precision is low, e.g. by no means could go beyond 0.5 micron. Such a precision can be realized in a bench-top system on top of an optical table, where the mechanical precision can be very good and vibrations are minimized during imaging procedure. However, for the purpose of future development of a rotary-gantry DPC-CBCT system, such a precision is very difficult to achieve at a reasonable cost.

To relax the requirement of mechanical precision, we propose a source grating stepping method to retrieve the phase information for a DPC-CBCT system using a hospital-grade x-ray tube coupled with a source grating. In order to acquire the N step images, we propose to shift the source grating N times, each $1/N$ of the source grating's period. The same physics and mathematics are used to retrieve the differentiated phase image. By shifting the source grating, which enjoys a much larger period than the self image, the requirement of mechanical precision can be greatly loosened.

This approach was validated using the bench-top DPC-CBCT system built in the Cone Beam CT Imaging Lab at the University of Rochester. A cylinder phantom was used in this research and the result indicated that source grating stepping approach is capable of producing comparable image quality as that acquired by phase grating/analyzer grating stepping approach.

The next section describes the phase stepping approach and the experiment setup to implement the phase stepping scheme. Section 3 shows preliminary DPC-CBCT reconstructions and quantitatively compares the results of source grating stepping (without error and with random error) and phase grating stepping. Section 4 concludes this research.

2. MATERIALS AND METHODS

2.1 The phase stepping approach

The principle of differential phase-contrast imaging using phase stepping approach has been well documented elsewhere [2–3] and it is briefly reviewed in this manuscript. A general hospital-grade x-ray tube cannot be used for phase contrast imaging because of its large focal spot size and thus very short coherence length. However, the DPC imaging technique is able to produce one-dimensional spatial coherence by applying an absorption grating (the

source grating G0) to a hospital-grade x-ray tube. The gold line patterns of G0 can absorb almost all x-ray photons impinging on them while the grooves in between let all the x-ray photons pass through. Thus the source grating divides a large focal spot x-ray source into several narrow line sources. Each of these line sources is able to produce sufficient spatial coherence at the direction perpendicular to the lines while they are mutually incoherent. When proper parameters are chosen, these line sources contribute constructively in the imaging process. The phase grating and the analyzer grating work as a shearing interferometer. The phase grating shows negligible absorption but substantial phase shift, dividing the x-ray beam into two first diffraction orders. The refracted beams then interfere and form periodical fringes at a fractional Talbot distance where the analyzer grating is placed. The period of the analyzer grating is chosen the same as the period of fringes, which is half of the period of the phase grating at the fractional Talbot distance. If the incident x-ray beam encounter an object before it reaches the phase grating, its wavefront will be perturbed by the object, leading to local displacement of the fringes. To retrieve the encoded phase information, the analyzer grating is then shifted within one grating period at several steps, and at each step an intensity image is recorded using an x-ray detector that has a pitch larger than the fringe period. Thus the recorded intensity images shows no fringe patterns but they are modulated based on the position of the analyzer grating, and the DPC image can be calculated using the phase stepping algorithm from these intensity images. A DPC image is contrast enhanced along the direction perpendicular to grating grooves as theoretically it shows the first derivative of phase shift along this direction. So the phase information is firstly encoded into the local displacement of fringe patterns, and then translated into the detected intensity images when being captured by the detector. The phase stepping algorithm is used to retrieve the phase information from these intensity images. Thus the contrast nature of the DPC image is different from that of an absorption image [6].

As shown in Figure 2, given the distance between source grating and phase grating R_1 and the distance between the phase grating and analyzer grating R_2 , the period of source grating p_0 and the analyzer grating p_2 , and period of the source grating could be calculated with the formula:

$$p_0 = p_2 \times \frac{R_1}{R_2} \quad (1)$$

It could be observed that the same phase grating stepping effect can be achieved by stepping the source grating. In our case, $p_2 = 4.0$ micron and $p_0 = 30.7$ micron, the travelling distance of the linear stage in phase grating stepping case is 0.5 micron for eight steps, while in the case of source grating stepping, each step equals to 3.83 micron, which is over seven times larger. With a longer travelling distance, naturally a higher tolerance of error could be achieved.

2.2 The DPC-CBCT system at our lab

A bench-top differential phase-contrast cone beam CT system has been designed and built in our laboratory for this study. The x-ray generating system consists of a hospital-grade x-ray tube (Varian Rad-70D, Varian X-ray Products, Salt Lake City, UT) and a high-frequency x-

ray generator (SHF1635MAMMO-480VAC, Sedecal USA, Arlington Heights, IL). The camera system includes a CMOS detector (RadEye HR (High Resolution) detector, Radicon Imaging Corp, Santa Clara, CA) and a frame grabber (PXD 1000, Imagenation, Beaverton, OR). An interface board together with software control, was developed to control the timing of x-ray shot and image acquisition. The phantom is mounted and rotated on a tomography sample stage (B4836TS Stage, Velmex Inc. Bloomfield, NY).

Gratings were fabricated at Cornell NanoScale Facility with an effective area of $60\text{mm} \times 60\text{mm}$ on 4-inch $\langle 110 \rangle$ silicon wafers. Gratings and system are optimized for an effective x-ray energy of 24 keV (wavelength of $0.518 \text{ \AA}$). Grating periods are $30.7 \text{ }\mu\text{m}$, $8.0 \text{ }\mu\text{m}$ and $4.6 \text{ }\mu\text{m}$ respectively for the source grating, phase grating and analyzer grating. The duty cycle is 50% for both phase grating and analyzer grating. Source grating opening width is $18 \text{ }\mu\text{m}$, and the distance between source grating to phase grating is 118 cm, which produces a coherence length of $3.4 \text{ }\mu\text{m}$ in the horizontal direction. The first fractional Talbot distance (17.72cm) is used in the setup. Detailed grating fabrication technology is referred to in references [7–8]. A motorized high precision linear stage (VP-25XA, Newport Research, Irvine, CA) is used to shift one of the gratings to generate phase shift with a step accuracy up to 0.1 micron.

Figure 3 shows the two system setups for source grating stepping and phase grating stepping. The red circle indicate the location of the linear stage for each case. Detailed system parameters are listed in Table 1.

2.3 Experiment design

To investigate the performance of the source grating stepping method, a phantom study was thus conducted. A cylinder phantom was used which was composed of a cylindrical acrylic wall with a diameter of 25.4mm and four solid material rods with a diameter of 6mm immersed in water. Using the identical system setting, (100mA, 320ms, 40kVp, 120 projection, 8steps), two sets of DPC images of the same cylinder phantom, were acquired via the phase grating stepping method and the source grating stepping method. Reconstructions were performed via the Hilbert filter-based FBP algorithm. The reconstruction results of both methods are compared in terms of background uniformity, noise and contrast.

In order to validate the source grating stepping method's contribution to mechanical stability, we deliberately assign random error to the position of each phase step while shifting the source grating. The random error ranges between -0.5 to 0.5 micron around the exact position. It should be noticed that as 0.5 micron equals to the length of each step in the phase grating/analyzer grating stepping scheme, such error during the scan would yield the phase stepping method powerless for this scheme. The result of such test is again compared with the phase grating stepping method and source grating stepping method without error case.

3. RESULTS

Reconstructions of a typical axial slice of the cylinder phantom are shown in Figure 5, results including the phase grating stepping approach 5(a), the source grating stepping approach 5(b) and source grating stepping with a random error in the range of ± 0.5 micron for each step. Measurements of background uniformity, noise level, and contrast of all the four material rods are listed in Table 2. It is clear that the results of source grating stepping approach is comparable to that of the phase grating stepping approach. And as expected, the source grating stepping method is shown to be able to stand up to ± 0.5 micron error as the image quality with random errors in steps is very close to that without random errors.

4. CONCLUSION

To improve the tolerance of mechanical errors, we investigated the method of source grating stepping in DPC-CBCT imaging. The bench top DPC-CBCT system enjoys much higher mechanical stability with source grating stepping due to 7x large step size. It has been demonstrated that the source grating stepping method could withstand up to, though not limited to, a random error of ± 0.5 micron in each step, which equals to the whole step of phase grating stepping approach. The measured image quality from the source grating stepping approach is comparable to that from the phase grating stepping. And we believe it can be improved with better source grating uniformity and system alignment. The result of this research greatly relaxed the mechanical precision required for a DPC-CBCT system, which will directly benefit future development of rotary-gantry system, where a high mechanical precision as that on an optical table cannot be achieved for a rotating imaging chain.

Acknowledgments

This project was supported in part by NIH Grants R01 CA 143050. David Conover is an employee of Koning Corporation. Ruola Ning is a consultant to, equity holder and President of Koning Corporation during the study. Koning Corporation has licensed several patents from the University of Rochester, and seeks to commercialize medical imaging equipment utilizing the cone beam CT technology. The University of Rochester holds a small equity interest in Koning. Part of this work was performed at the Cornell NanoScale Facility (CNF), a member of the National Nanotechnology Infrastructure Network, which is supported by the National Science Foundation (Grant ECS 0335765).

References

1. Momose A, Kawamoto S, Koyama I, Hamaishi Y, Takai K, Suzuki Y. Demonstration of x-ray Talbot interferometry. *Jpn J Appl Phys.* 2003; 42:L866–L868.
2. Weitkamp T, Diaz A, David C, Pfeiffer F, Stampanoni M, Cloetens P, Ziegler E. X-ray phase imaging with a grating interferometer. *Opt Express.* 2005; 13(16):6296–6304. [PubMed: 19498642]
3. Pfeiffer F, Weitkamp T, Bunk O, David C. Phase retrieval and differential phase-contrast imaging with low-brilliance x-ray sources. *Nature Physics.* 2006; 2:258–261.
4. Qi Z, Zambelli J, Bevins N, Chen G. Quantitative imaging of electron density and effective atomic number using phase contrast CT. *Phys Med Biol.* 2010; 55:2669–2677. [PubMed: 20400806]
5. Yu Y, Ning R, Cai W. Performance evaluation of a differential phase-contrast cone-beam CT (DPC-CBCT) system for soft tissue imaging. *Proc SPIE.* 2011; 7961:79614X.
6. Cai W, Ning R. Dose efficiency consideration for volume-of-interest breast imaging using x-ray differential phase-contrast CT. *Proc SPIE.* 2009; 7258:72584D.

7. Yu Y, Ning R, Cai W. Performance evaluation of a differential phase-contrast cone-beam CT (DPC-CBCT) system for soft tissue imaging. *SPIE Proc Physics of Medical Imaging*. 2011; 7961:79614X.
8. David C, Bruder J, Rohbeck T, Grunzweig C, Kottler C, Diaz A, Bunk O, Pfeiffer F. Fabrication of diffraction gratings for hard x-ray phase contrast imaging. *Microelectronic Eng*. 2007; 84:1172–1177.

Author Manuscript

Author Manuscript

Author Manuscript

Author Manuscript

Figure 1.
Illustration of a bench-top differential phase-contrast cone beam CT system.

Figure 2.
Geometrical requirements of grating periods

Figure 3. Experimental system setups using phase stepping approach for differential phase-contrast cone beam CT. Upper: source grating stepping and Lower: phase grating stepping

Figure 4.

The cylinder phantom used in this study. It is composed of a cylindrical acrylic wall with a diameter of 25.4mm and four solid material rods with a diameter of 6mm immersed in water.

Figure 5. Reconstruction slices of phase grating stepping (left), source grating stepping (middle) and source grating stepping with ± 0.5 micron error (right).

Table 1

Major parameters of the DPC-CBCT system

Focal spot size	0.3 mm
Detector pixel size	22.5 μm for 1x1 rebin 45.0 μm for 2.2 rebin
Source-to-phase-grating distance	118 cm
Phase-to-analyzer-grating distance	17.72 cm
Object-to-detector distance	25.4 mm
Field of view	32 mm $\times$ 24 mm
Cone angle	$< 2^\circ$
Tube peak voltage	25–49.5 kVp
Tube current	16–200 mA
Exposure time per exposure	320 ms
Projection number	120
Recon voxel size	(36.8 μm) ³
Phase grating period	8.0 micron
Analyzer grating period	4.6 micron
Source grating period	30.7 micron

Author Manuscript

Author Manuscript

Author Manuscript

Author Manuscript

Table 2

Image quality comparison in terms of uniformity, noise and contrast.

	Background Uniformity	Background Noise Level	Contrast PI	Contrast PC	Contrast PTFE	Contrast PS
Phase Grating Stepping	15.2%	8.1%	+90.8%	+13.4%	38.9%	0.0%
Source Grating Stepping	18.5%	10.5%	+70.1%	+11.9%	+34.4%	0.0%
Source Grating Stepping (with error)	19.0%	10.9%	+68.4%	+11.7%	+35.2%	0.0%