52

X-ray Phase-Contrast Tomosynthesis Imaging

Ke Li and Guang-Hong Chen

CONTENTS
52.1 Introduction .................................................................................................................. 1049
52.2 Principles ...................................................................................................................... 1050
52.3 Image Acquisition System .......................................................................................... 1051
52.4 Image Reconstruction Methods .................................................................................. 1052
  52.4.1 Shift-and-Add ......................................................................................................... 1052
  52.4.2 Filtered Back Projection (FBP) .............................................................................. 1052
52.5 Image Quality .............................................................................................................. 1054
  52.5.1 Mean Signal Value ............................................................................................... 1054
  52.5.2 Noise .................................................................................................................. 1055
  52.5.3 Contrast-to-Noise Ratio (CNR) .......................................................................... 1055
  52.5.4 Spatial Resolution ............................................................................................... 1057
52.6 Image Artifacts ............................................................................................................ 1058
  52.6.1 Out-of-Plane Signal Leakage .............................................................................. 1058
  52.6.2 Stripe Artifacts ................................................................................................... 1059
  52.6.3 Truncation Artifacts ............................................................................................ 1060
52.7 Future Prospects ......................................................................................................... 1060
References ......................................................................................................................... 1062

52.1 Introduction

X-ray tomosynthesis is a pseudo-tomographic imaging method that can partially address the problem of structural overlap encountered in planar X-ray imaging (e.g., rib-lung overlap in chest X-ray, or normal breast tissue-focal lesion overlap in mammography) (see Section II, Chapter 20). In a typical X-ray tomosynthesis image acquisition, sequential two-dimensional (2D) projection images are recorded at different view angles within a limited angular span (e.g., ±30°), and the projection image dataset is processed with a limited-angle tomographic reconstruction algorithm to synthesize three-dimensional (3D) images of the image object (Figure 52.1). These tomosynthesis images can be considered as a stack of in-focus planes cutting through the object at different depths, thus alleviating the structural overlap issue. Compared with X-ray computed tomography (CT) that can record hundreds of projections over 360° in less than a second (see Section III, Chapter 32), the total angular range, the image acquisition speed, and the detector readout speed are greatly reduced in tomosynthesis imaging, which helps to lower its system cost. In addition, a single tomosynthesis image acquisition does not require multiple full tube rotations as in modern CT, and the detector can be kept stationary (see Section II, Chapter 23). Therefore, a typical tomosynthesis imaging system does not require a costly slip-ring gantry, as in CT. Instead, a tomosynthesis capable gantry can be quite similar to a standard planar X-ray imaging system with the addition of limited angular motion. Another major benefit of tomosynthesis is that the flat panel detectors used in tomosynthesis systems typically have spatial resolution about an order of magnitude better than a typical CT detector, leading to very high in-plane spatial resolution. On the other hand, due to the limited angular range of tomosynthesis, its spatial resolution along the depth (z) direction is generally inferior to that of true 3D imaging such as CT. As a result, tomosynthesis images are usually reconstructed with small in-plane pixel size (e.g., 100 μm), but a much thicker slice (e.g., 1 mm).

In recent years, X-ray tomosynthesis has attracted considerable attention, especially in the field of breast imaging (Niklason et al. 1997; Suryanarayanan et al. 2000; Bissonnette et al. 2005; Ren et al. 2005; Smith 2005; Andersson et al. 2008; Sechopoulos 2013a,b; Vedantham et al. 2015). Several digital breast tomosynthesis (DBT) imaging systems have been FDA-approved and employed in clinical practice (Feng and Sechopoulos 2012). Multiple clinical trials have demonstrated that, compared with digital mammography (DM), DBT led to a substantial increase in the sensitivity of breast cancer detection (Andersson et al. 2008; Ciatto et al. 2013; Skaane et al. 2013; Houssami et al. 2014), although it also increased false positive rates and recall rates (Lang et al. 2016). X-ray tomosynthesis imaging has also been applied to lung imaging (Vikgren et al. 2008; Dobbins and McAdams 2009; Gomi et al. 2013), cardiac interventional imaging (Nett et al. 2008; Speidel et al. 2010), musculoskeletal
All of these tomosynthesis imaging applications share a common feature: they all utilize the absorption of X-rays to generate image contrast. However, absorption is not the sole contrast mechanism enabled by X-ray-matter interactions: when X-rays propagate through an object, both X-ray absorption (a particle physics phenomenon) and X-ray phase shift (a wave optics phenomenon) occur, and, in principle, both effects can be utilized to generate image contrast (see Section IV, Chapter 49). Interestingly, these two contrast mechanisms can be related through the complex refractive index $n = 1 - \delta + i\beta$, as the real component is related to phase shift, while the imaginary component is related to the well known X-ray linear attenuation coefficient, $\mu$, as follows: $\beta = \lambda\mu/4\pi$, where $\lambda$ is the wavelength of the X-ray. In the diagnostic X-ray energy range, the difference in $\mu$ between two materials is primarily determined by the difference in their effective atomic number ($Z$), and, for soft issues with similar $Z$, their absorption contrast is very limited. The primary motivation of introducing the phase-contrast mechanism to X-ray tomosynthesis imaging is the potential in improving soft tissue contrast sensitivity, as $\delta$ is determined by the electron density of the material and is independent of $Z$.

So far, several phase-contrast tomosynthesis imaging (PCTI) methods have been developed, including a grating-based method (Li et al. 2012, 2013b, 2014; Schleede et al. 2014; Garrett et al. 2015), a free propagation or in-line holography-based method (Zhang et al. 2010; Hammonds et al. 2011, 2013; Wu et al. 2015), and a crystal diffraction enhanced imaging (DEI)-based method (Maksimenko et al. 2007; Kang et al. 2008). This chapter will cover the basic principles of PCTI, its data acquisition system, reconstruction algorithms, image quality assessments, and image artifacts. In each of these topics, PCTI will be compared with conventional absorption contrast tomosynthesis imaging (ACTI). The chapter concludes with an outlook to the future research directions of PCTI. Most of the examples provided in this chapter were obtained using a grating-based PCTI system in the authors’ laboratory, but the scope of the chapter is not limited to the grating-based method.

### 52.2 Principles

In the diagnostic X-ray energy range, the accumulated phase shift (relative to vacuum) for an X-ray passing through an image object is given by

$$\phi = -\frac{2\pi}{\lambda}\int \delta(x) \, dl,$$

where $l$ is along the X-ray propagation direction, and $\lambda$ is the wavelength of the X-ray. For grating-based PCTI systems, a projection image recorded at view angle $\theta$ is related to the first-order derivative of $\phi$ by (Momose et al. 2003; Weitkamp et al. 2005; Pfeiffer et al. 2006):

$$\varphi(u,v,\theta) = \frac{\lambda d}{p} \nabla_{\theta} \phi = -2\pi d \frac{\partial}{\partial u} \int_{0}^{\theta} \delta(x) \, dl,$$

where $d$ is the distance between a diffraction grating and the X-ray detector, and $p$ is the period of the X-ray diffraction pattern at the detector plane, $(u,v)$ represents a 2D Cartesian coordinate system defined in the detector plane, with the $u$ axis being perpendicular to the periodic grating structure. Grating-based X-ray phase-contrast imaging is often referred to as X-ray differential phase-contrast imaging, since its image signal is related to the spatial derivative of the X-ray phase shift $\phi$. In comparison, the projection image of ACTI is related to the real part of the refractive index by...
X-ray Phase-Contrast Tomosynthesis Imaging

1051

FIGURE 52.2 (a) The Grossman tomosynthesis image acquisition geometry. (b) The DBT image acquisition geometry. (From Li, K. et al. 2014. Medical Physics 41:011903, with permission.)

\[
R(u,v,\theta) = \log \left( \frac{I_1}{I_0} \right) = \int_{\delta} \mu(\delta) \, d\delta \tag{52.3}
\]

where \( I_0 \) and \( I_1 \) are the measured photon number with and without the image object. Similar to grating-based PCTI systems, the projection image of DEI-based PCTI systems is also linearly proportional to the first-order derivative of \( \phi \). In comparison, the projection signal of an in-line PCTI system is related to the Laplacian of \( \phi \) by (Wilkins et al. 1996):

\[
I(u,v,\theta) = I_0 \exp \left\{ -\int_{\delta} \mu(\delta) \, d\delta \right\} \left[ 1 + \frac{\lambda d}{2\pi} \nabla^2_{\tilde{\gamma} \tilde{\gamma}} \phi \right] \tag{52.4}
\]

where \( d \) is the distance between the image object and the detector. The Laplacian term captures boundaries and interfaces with abrupt changes in \( \phi \), and the magnitude of edge enhancement increases with larger object-to-detector distance. Since both \( \mu \) and \( \phi \) contribute to the in-line image signal, additional phase retrieval is needed to separate the phase-contrast signal from the absorption contrast signal, otherwise pixel values of the reconstructed tomosynthesis images are not linearly related to \( \delta \).

As in phase-contrast CT (PCCT), PCTI is able to reconstruct a volumetric distribution of \( \phi \) from multiple phase-contrast projection images measured at different view angles. However, unlike PCCT, PCTI measures projection images only within a limited angular span (e.g., from \(-30^\circ \) to \(+30^\circ \)), therefore it produces a “distorted” version of \( \delta(\tilde{\gamma}) \), particularly along the interface relevant to the image object. To measure multiple projection images over a finite angular span, three system geometries can be potentially used. The first, shown in Figure 52.2a, is known as the Grossman geometry (Grossman 1935), in which both the X-ray source and detector rotate around the image object. The second, shown in Figure 52.2b, is commonly used in state-of-the-art clinical DBT systems. In this geometry, the X-ray source rocks along a circular trajectory in the \( x-z \) plane, while both the image object and the detector are kept stationary. If a grating-based PCTI system uses this geometry, the orientation of a 1D structure in the gratings should be parallel to the \( x-z \) plane, otherwise the X-ray beam could be partially blocked by the sidewalls of the grating structure (Figure 52.3). In comparison, the gratings can take any orientation within the \( u-v \) plane in the Grossman geometry, since the X-ray beam is always perpendicular to the grating surface. The third geometry is the inverse version of the Grossman geometry, in which the positions of the tube, detectors, gratings, or crystals are fixed, and the image object rotates during the PCTI acquisition. This inverse Grossman geometry has been widely used in published work on PCTI, and it is the only feasible geometry in DEI-based PCTI systems that require the use of synchrotron radiation (Maksimenko et al. 2007; Kang et al. 2008).

52.3 Image Acquisition System

To date, several experimental PCTI acquisition systems have been reported in the literature. The earliest system was based on the DEI method and used a synchrotron X-ray source, crystal

FIGURE 52.3 (a) For grating-based PCTI systems, when the 1D structure in the grating is aligned with the \( y \) axis, X-rays may be partially blocked by the wall of the grating at an oblique incident angle. (b) When the grating structure is aligned with the \( x \) axis, it is parallel to the plane in which the tube rocks; therefore, this orientation prevents incident X-rays from being blocked by the grating at any oblique projection angles. This setup also increases the effective thickness of the X-ray absorbing layer in the grating, which may translate into higher fringe visibility.
monochromator, and crystal analyzer (Maksimenko et al. 2007; Kang et al. 2009). The first non-synchrotron-based PCTI system was reported in Hammonds et al. (2011), and used a microfocus laboratory tube with a fixed tungsten anode. The first grating-based PCTI system was reported in Li et al. (2012). A hospital-grade rotating anode diagnostic X-ray tube with 1.0 mm nominal focal spot size was used in this system, which can significantly shorten the image acquisition time compared with microfocus and fixed-anode tubes. All systems reported so far use the inverse Grossman geometry, and the total angular span ranges from ±8° to ±30°. Table 52.1 summarizes the major parameters of these PCTI systems.

### 52.4 Image Reconstruction Methods

#### 52.4.1 Shift-and-Add

The shift-and-add (SAA) method, also known as the direct back projection method, is the most straightforward way to reduce structural overlap in planar projection images and reconstruct tomosynthesis images (Niklason et al. 1997). In SAA, pixels in each projection are shifted along the x direction by a certain distance, and then all the projections are added together to enhance in-focus features and cancel out-of-focus features. The amount of lateral shift depends on the acquisition geometry; for systems using the Grossman or inverse Grossman geometry, the formula for SAA-based PCTI reconstruction is given by (Li et al. 2014):

\[
I_{\text{PC}}(x,y,z) = \Delta \theta \sum_{\theta} \phi(x \cos \theta + z \sin \theta, y, \theta),
\]

where \( I_{\text{PC}} \) denotes the pixel value of phase-contrast tomosynthesis image volume, and \( \Delta \theta \) is the angular interval between two consecutive view angles. The SAA formula for systems with the DBT geometry can be found in Niklason et al. (1997) and Wu et al. (2004).

Figure 52.4 shows phase-contrast tomosynthesis images acquired with a grating-based system and reconstructed using SAA. Compared with the planar differential phase-contrast (DPC) projection image, the overlap of features inside a cylindrical water phantom was effectively reduced, and the conspicuity of these features were improved. However, the differential appearance of the DPC projection image was inherited by the reconstructed tomosynthesis images. In other words, the pixel values of SAA-based phase-contrast tomosynthesis images are not linearly related to the volumetric distribution of \( \delta \). This is also true for in-line and DEI-based PCTI systems. Hammonds et al. (2013) used an attenuation-partition based algorithm (APBA) to retrieve phase shift, \( \phi \), from in-line phase-contrast projection images, then used SAA to reconstruct PCTI. The pixel value of the reconstructed images was found to be linearly proportional to \( \phi \), although it is unclear whether the value is also linearly related to \( \delta \).

#### 52.4.2 Filtered Back Projection (FBP)

For grating and DEI-based PCTI systems, FBP can be used to remove the differential appearance of phase-contrast tomosynthesis images and achieve semi-quantitative imaging of \( \delta \). To derive the FBP reconstruction algorithm for PCTI, let’s first revisit the FBP algorithm for ACTI (see Section III, Chapter 33). It is well known that a ramp kernel can be used to filter the absorption contrast projection data to help reconstruct the volumetric distribution of \( \mu \). In the frequency space, this filtering operation can be described as (Kak and Slaney 1988; Lauritsch and Harer 1998):

\[
\hat{R}'(f_x, f_y, \theta) = \hat{R}(f_x, f_y, \theta) \cdot |f_y|,
\]

where \(|f_y|\) is the ramp kernel, and \( \hat{R}(f_x, f_y, \theta) \) denotes the 1D Fourier transform of \( R \) with respect to the \( u \) axis, namely:

\[
\hat{R}(f_x, f_y, \theta) = \int R(u, f_y, \theta) \ e^{-2\pi i u f_y} \ du.
\]

Similarly, we can write the 1D Fourier transform of the phase-contrast projection image as follows (Li et al. 2014):

\[
\hat{I}_{\text{PC}}(f_x, f_y, \theta) = \int I_{\text{PC}}(x, f_y, \theta) \ e^{-2\pi i u x} \ dx.
\]
X-ray Phase-Contrast Tomosynthesis Imaging

where $\bar{R}_\delta$ denotes the line integral of $\delta$. Based on Equation 52.8, the following operation is equivalent to filtering $\bar{R}_\delta$ with a ramp kernel:

$$
\bar{R}_\delta \cdot [f] = \left[ - \frac{p}{2 \pi d} \frac{1}{12 \pi f_a} \right] \varphi(f_a, v, \theta) \cdot [f] \tag{52.9}
$$

Therefore, the measured DPC projection data need to be filtered by a modified Hilbert kernel rather than the ramp kernel to effectively remove the differentiation operator and quantitatively restore the volumetric distribution of $\delta$. In the spatial domain, the filtering can be expressed as the following convolution operation:

$$
\phi'(u,v,\theta) = \phi(u,v,\theta) \otimes \frac{p}{4 \pi^2 d} \delta(u', v, \theta) \tag{52.10}
$$

where $1/4\pi^2d$ is the Fourier pair of the modified Hilbert kernel $i(p/4\pi^2d)\text{sgn}(f_a)$. The filtering is performed separately for each projection view angle $\theta$, then the filtered projections can be back projected to reconstruct phase-contrast tomosynthesis images.

For systems with the Grossman geometry and under the parallel beam approximation, the back projection can be analytically described as follows:

$$
I_{PC}(x,y,z) = \sum_{\theta} \phi'(x\cos\theta + z\sin\theta, y, \theta) \tag{52.11}
$$

where $\phi'(u,v,\theta) = \phi(u,v,\theta) \otimes \frac{p}{4 \pi^2 d} \delta(u', v, \theta)$.

Figure 52.5 compares phase-contrast tomosynthesis images reconstructed by SAA and FBP. A grating-based system was used for data acquisition. The FBP algorithm removed the differential appearance of the image. The negative overshooting at the left and right boundaries of the object in the FBP image was caused by the incomplete angular coverage and is commonly observed in tomosynthesis imaging (Sechopoulos 2013b). Figure 52.6 shows another example of phase-contrast tomosynthesis images reconstructed by FBP.

For systems with divergent beams, Equation 52.11 can be modified based on the Feldkamp–Davis–Kress (FDK) algorithm to reconstruct phase-contrast tomosynthesis images (Wu et al. 2015). The back projection can also be implemented using projection matrices, particularly for non-Grossman systems and/or systems with divergent X-ray beams (Galigekere et al. 2003). It has been demonstrated that the pixel value of phase-contrast tomosynthesis images reconstructed using the FBP method is linearly proportional to $\delta$ (Li et al. 2014).
52.5 Image Quality

In general, methods and metrics for evaluating the image quality of PCTI are similar to those of ACTI, and can be divided into the following categories: signal, noise, signal normalized by noise, and spatial resolution (see Section I, Chapter 15). Image artifacts will be discussed in a separate section later in this chapter. Due to the nonconventional differential appearance in some phase-contrast tomosynthesis images, particularly those reconstructed by SAA, the assessment of their image quality may also use some special metrics, which will be described in this section.

52.5.1 Mean Signal Value

When SAA is used for tomosynthesis reconstruction, the differential appearance in the projection data is inherited by the reconstructed phase-contrast tomosynthesis images, therefore it is difficult to measure the mean signal value of an image object, unless it is a ramp with constant slope (Ge et al. 2016). In comparison, the mean signal value can be measured in phase-contrast tomosynthesis images reconstructed with FBP, since the differential appearance is removed by the use of the Hilbert kernel. The measurement is usually performed in a region that corresponds to a uniform region in a relatively large object, such as the phantom inserts shown in Figure 52.5. It has been demonstrated in Li et al. (2014) that the mean signal value of FBP-based phase-contrast (PC) tomosynthesis images is linearly proportional to $\delta$, and it is quantitatively related to that of FBP-based absorption contrast (AC) tomosynthesis images generated from the same data acquisition by (Li et al. 2014):

$$\frac{I_{PC}^{FBP}}{I_{AC}^{FBP}} = \frac{\delta}{\mu}. \tag{52.12}$$
In other words, the mean signal magnitude of absorption contrast tomosynthesis images can be used to quantitatively predict that of phase-contrast tomosynthesis images, and vice-versa. This is of great potential value for the design and optimization of PCTI systems.

By separating phase shift information from the absorption signal using the attenuation-partition based algorithm (APBA), Hammonds et al. (2013) validated the signal linearity of phase-contrast tomosynthesis images generated with an in-line system. They showed that the retrieved phase signal of polystyrene plates was proportional to the total thickness of the plates.

### 52.5.2 Noise

The noise performance of PCTI systems can be quantified by the noise standard deviation (σ), which is a zero-frequency metric that quantifies the magnitude of noise. Similar to the mean signal value, the noise standard deviation is usually measured in a uniform region in the reconstructed tomosynthesis image. Alternatively, it can be measured from noise-only images obtained by subtracting image volumes generated by two back-to-back acquisitions (Li et al. 2014). Wu et al. (2015) measured noise standard deviation of high energy (120 kVp) in-line tomosynthesis images and found about 20% reduction in σ compared with conventional absorption contrast tomosynthesis.

In addition to noise standard deviation, NPS is another quantitative metric to characterize the noise performance of PCTI systems. Unlike noise standard deviation, NPS is a frequency-dependent metric that quantifies both the magnitude and spatial correlation of noise. Since tomosynthesis is a volumetric imaging method, it is usually preferable to use the three-dimensional (3D) NPS to capture the noise correlation both in the axial (x–y) plane and along the z direction. The 3D NPS of phase-contrast tomosynthesis images can be calculated as (Li et al. 2014):

\[
\text{NPS}^\text{3D}_{\text{nc}} = \frac{\Delta I_{xy z}}{N_x N_y N_z} \left| \text{DFT}_\text{3D} \left( \Delta \text{VOI}_{\text{nc}} \right) \right|^2,
\]  

(52.13)

where DFT$_{3D}$ denotes 3D discrete Fourier transform, $\Delta \text{VOI}_{\text{nc}}$ is a volume of interest (VOI) extracted from the 3D phase-contrast tomosynthesis images and detrended by its background, $\Delta x$ and $N_x$ denote the pixel size and number of pixels of $\Delta \text{VOI}_{\text{nc}}$ along the x axis, respectively, and $\langle \cdot \rangle$ denotes an ensemble average of multiple VOIs obtained by either shifting the location of the VOI within a PCTI volume or repeated PCTI acquisitions. Using the same methodology, the 3D NPS of absorption contrast tomosynthesis images can also be measured and compared with that of PCTI.

Figure 52.7 shows 3D NPS of PCTI and ACTI measured using a cylindrical phantom (Li et al. 2014). A total of 121 VOIs were obtained by sliding an $N_y = N_z = 180 \times 180 \times 30$ volumetric window within the 3D tomosynthesis image volume of the phantom. For phase-contrast tomosynthesis images reconstructed by SAA and FBP, their NPSs show no fundamental difference except their magnitudes, because the modulus of the Hilbert kernel used in FBP is frequency independent. Compared with absorption contrast tomosynthesis images reconstructed using FBP and the ramp kernel, the phase-contrast tomosynthesis images contain relatively less high frequency noise and less aliased noise. Instead, they are dominated by low- and mid-frequency noise. In particular, the $f_x$-$f_z$ plane of the 3D NPS looks like an hourglass, and the two missing cones are caused by the limited view angle coverage in PCTI acquisition.

Figure 52.7 also shows that the magnitudes of the 3D NPS of PCTI reconstructed by SAA and FBP differ by about 10 orders of magnitude. This difference in noise magnitudes is introduced by the extra $4\pi^2 d/\rho$ factor in the modified Hilbert kernel used in FBP, whereas the SAA reconstruction does not use this scaling factor.

Li et al. (2013a,b) performed a cascaded systems analysis of the noise transfer process in the grating-based PCTI system and found that the 3D NPS of PCTI is quantitatively related to that of ACTI. When SAA is used for tomosynthesis image reconstruction, the 3D NPS of PCTI and ACTI are related by:

\[
\text{NPS}^\text{3D}_{\text{nc}} (f_x, f_y, f_z) = \frac{2}{\varepsilon^2} \text{NPS}^\text{3D}_{\text{nc}} (f_z, f_y, f_z),
\]

(52.14)

where $\varepsilon$ is the fringe visibility of the interference pattern. When SAA is used for tomosynthesis image reconstruction, the 3D noise power spectrum (NPS) of PCTI and ACTI are related by (Li et al. 2013a,b):

\[
\text{NPS}^\text{3D}_{\text{nc}} (f_x, f_y, f_z) = \frac{2p^2}{(4\pi^2 d \rho)^2} \frac{1}{f_x^2 + f_z^2} \text{NPS}^\text{3D}_{\text{ac}} (f_x, f_y, f_z). \quad (52.15)
\]

The scaling factor of $(f_y^2 + f_z^2)^{-1}$ leads to the differences in the shape of the 3D NPS between the two contrast mechanisms, and the numerical scaling factor $2p^2/(4\pi^2 d \rho)^2$ is responsible for the difference in noise magnitude. Figure 52.8 shows a comparison between the measured and predicted NPS of PCTI.

The NPS of the PCTI system also depends on the specific phase-contrast imaging method. As an example, the NPS of in-line and grating-based PCTI systems are different. Due to the use of ramp (instead of Hilbert) kernel, the NPS of an in-line PCTI system has a similar shape to that of a conventional ACTI system and has a peak at a certain intermediate frequency along the $f_y$ axis (Wu et al. 2015).

### 52.5.3 Contrast-to-Noise Ratio (CNR)

In tomosynthesis imaging, CNR is also referred to as the signal difference-to-noise ratio (SDNR). CNR or SDNR can be calculated as

\[
\text{CNR} = \frac{\tilde{T}_{\text{object}} - \tilde{T}_{\text{bg}}}{(\sigma_{\text{object}} + \sigma_{\text{bg}})/2},
\]

(52.16)

or alternatively,

\[
\text{CNR} = \frac{|\tilde{T}_{\text{object}} - \tilde{T}_{\text{bg}}|}{\sqrt{(\sigma_{\text{object}}^2 + \sigma_{\text{bg}}^2)/2}}.
\]

(52.17)

where $\tilde{T}_{\text{object}}$ and $\tilde{T}_{\text{bg}}$ are the mean signal value of the image object and the background, respectively, in the tomosynthesis images.
FIGURE 52.7  Comparison of 3D NPS of PCTI and ACTI. (a) SAA-based PCTI. (b) FBP-based PCTI. (c) FBP-based ACTI. The first three rows are generated by projecting the 3D NPS along the y, x, and z axes, respectively. Numbers in the square brackets are the display ranges for each image. Line profiles through the third row (indicated by the arrows) are shown in the bottom row. (From Li, K. et al. 2014. Medical Physics 41:011903, with permission.)

FIGURE 52.8  Comparison of predicted (a) and measured (b) 3D NPS of FBP-reconstructed PCTI. The prediction was performed from the 3D NPS of FBP-reconstructed ACTI using Equation 52.15.
Li et al. (2014) measured the SDNR of phase-contrast tomosynthesis images of two physical phantoms acquired from a grating-based system. Compared with the associated ACTI, PCTI improved the SDNR of several material pairs such as acrylic-water and polystyrene (PS)-water (Figure 52.9), although ACTI generated higher SDNR for several other material combinations with high absorption contrast (e.g., Teflon-water).

Hammonds et al. (2013) measured the CNR of phase-contrast tomosynthesis images of an acrylic phantom acquired from an in-line system. Phase retrieval was performed prior to SAA image reconstruction. It was found that the CNR of acrylic (relative to water) is 25% higher in PCTI than in ACTI.

For an in-line PCTI system, a modified form of CNR can be used to quantify the improvement in the conspicuity of edges and boundaries. This modified CNR is referred to as the edge enhancement-to-noise ratio (EE/N); it uses the difference between the maximal and minimal intensity near an edge as the contrast. Hammonds et al. (2011) showed that, compared with planar phase-contrast radiographs, the EE/N of an air-fiberglass resin interface was improved in phase-contrast tomosynthesis images due to improved contrast and reduced noise. Wu et al. (2015) measured the CNR at the edges of air bubbles in a bubble wrap and showed that in-line PCTI resulted in a factor of 3 improvement in CNR compared with conventional ACTI.

### 52.5.4 Spatial Resolution

The in-plane spatial resolution of tomosynthesis images can be characterized using line profile through the edge of an object or with the modulation transfer function (MTF) (Samei et al. 2013). Figure 52.10 shows line profiles of a 1.9 mm Teflon sphere in PCTI and ACTI. A grating-based system was used for image acquisition, and FBP was used for tomosynthesis image reconstruction. Following normalization by the corresponding peak intensities, the profiles of PCTI and ACTI are matched along both the x and y direction, which suggests the in-plane spatial resolution of PCTI is equivalent to that of ACTI in grating-based systems. The negative edge overshoot along the x direction was caused by the incomplete angular coverage and was observed in both contrast mechanisms.

Wu et al. (2015) measured the MTF of their in-line PCTI system using a high contrast edge phantom and found that the MTF has a slight dependence on the X-ray energy. They further compared the line profiles of phase-contrast and absorption contrast tomosynthesis images of the edge phantom and found that in-line PCTI enhanced the sharpness of the edge in the axial tomosynthesis image plane.

Spatial resolution along the z direction, namely the effective slice thickness, can be characterized by plotting the line profile of a small and high contrast object along z. Figure 52.11 shows the z profile of a 1.9 mm Teflon sphere in DCTI and ACTI. A grating-based system was used for image acquisition, and FBP was used for tomosynthesis image reconstruction. For both contrast mechanisms, the full-width-at-half maximum (FWHM) of the z profile is 4.0 mm, indicating equivalent spatial resolution along the z direction.
For tomosynthesis imaging, spatial resolution along the \( z \) direction can also be quantified by the Artifact Spread Function (ASF) defined as follows:

\[
ASF(z) = \frac{T_{\text{object}}(z) - T_{\text{BG}}(z)}{T_{\text{object}}(z_0) - T_{\text{BG}}(z_0)},
\]

(52.18)

where \( z_0 \) is the location of the in-focus plane for the ASF test object, \( T_{\text{object}} \) is the mean pixel value of a region of interest (ROI) that corresponds to the location of the test object, and \( T_{\text{BG}} \) is the mean pixel value of the background region. In practice, ASF is often measured using a uniform sphere; however, the size and contrast of the sphere directly impact the width of the ASF along the \( z \)-axis. As a result, ASF is not an absolute metric and, when it is used to compare slice thickness across different tomosynthesis systems or reconstruction methods, the same test object should be used. Figure 52.12 shows PCTI and ACTI of a 6.3 mm polystyrene (PS) sphere suspended in vegetable oil (Li et al. 2014). Using this phantom, ASF was measured and the results are shown in Figure 52.13. This confirmed again that there is no fundamental difference in slice thickness between the two contrast mechanisms.

Spatial resolution along the \( z \)-axis has also been characterized for an in-line PCTI system in Hammonds et al. (2011). The characterization was performed using a 1-mm air hole inside a cylindrical phantom. The mean signal of the hole was plotted as a function of \( z \) position, and the FWHM of the plot was found to be 4-times the diameter of the hole.

## 52.6 Image Artifacts

### 52.6.1 Out-of-Plane Signal Leakage

The most common type of image artifact in tomosynthesis imaging is out-of-plane signal leakage, with phase-contrast tomosynthesis imaging being no exception (for a description of image artifacts in CT imaging, see Section III, Chapter 32). Figure 52.14 shows an example of this type of artifact: the signal of a 1.9 mm sphere is still observable in a plane 8 mm away from the in-plane. In tomosynthesis imaging, the out-of-plane signal leakage is caused by incomplete angular coverage: Based on the Tuy’s data consistency condition, the minimal angular span of projection image acquisition should be 180° + the fan angle, in
order to accurately reconstruct a tomographic image of the image object. The angular coverage of a tomosynthesis image acquisition is usually no more than 60°; therefore, a tomosynthesis image is essentially the non-ideal version of a true tomographic image. This point can be better illustrated in the “coronal” (x–z plane) reformats of the 3D tomosynthesis image volume shown in Figure 52.15. The limited angular coverage (±30°) of the tomosynthesis acquisition is clearly reflected by these reformats. A significant portion of the needed data is missing outside the “hourglass.” The shape of the sphere was decently restored along directions perpendicular to the projection directions, but was distorted in other directions.

52.6.2 Stripe Artifacts

In addition to signal leakage caused by incomplete angular coverage, tomosynthesis images may also contain in-plane stripe artifacts caused by an inadequate angular sampling rate: unlike in CT where the angular interval (Δθ) between two consecutive projections is less than 0.5°, in tomosynthesis imaging, the interval is usually 1° or larger. As shown in Figure 52.16, inadequate angular sampling will translate into streaks in the x–z plane, which correspond to stripes in the orthogonal x–y plane. This type of artifact is usually more prominent in planes far from the center of rotation, as the required angular sampling rate increases.
with the distance from the center of rotation (Barrett and Myers 2004). Figure 52.16 shows an example of the stripe artifact, which was found in both PCTI and ACTI. One method to reduce this artifact is to reconstruct thick tomosynthesis image slice, which can partially cancel the stripes in different $z$ positions.

### 52.6.3 Truncation Artifacts

In X-ray imaging, truncation is the cutoff of the signal from an image object beyond the maximal field-of-view (FOV) supported by the detector and X-ray beam. Absorption contrast tomosynthesis images reconstructed by FBP are particularly sensitive to truncation: When the projection data are filtered by a ramp kernel, additional spatial correlations will be introduced, which will amplify the abrupt signal change across the edge of the FOV caused by truncation. After back projection, the abrupt signal change will translate into the tomosynthesis image domain as stripe artifacts at the (left and right) peripheral regions in the image (Figure 52.17). For grating-based PCTI, truncation artifacts are usually less prominent due to the use of Hilbert kernel, which does not boost high frequency content in the projection data. However, PCTI is not immune to truncation. As shown in Figure 52.17, truncation artifacts manifest themselves as low frequency blooming and shading artifacts in the left and right peripheral regions of the phase-contrast tomosynthesis image.

To correct for truncation artifacts in PCTI, Garrett et al. (2015) investigated several extrapolation-based methods and compared their tomosynthesis-specific methods with traditional CT corrections. In their work, the projection data were extrapolated symmetrically by 50% of the detector width on each side along the scanning direction. This extent was chosen empirically to ensure projection data were available for all measured rays intersecting the reconstruction field of view (FOV). With extrapolation, they were able to mitigate truncation artifacts in PCTI images (see Figure 52.18). The authors also point out that truncation artifact correction method developed for ACTI led to residue artifacts in PCTI images, and vice-versa, and thus each contrast mechanism should be corrected using its own method. This difference in the optimal truncation correction method between PCTI and ACTI can be attributed to the difference in reconstruction kernel used in FBP.

### 52.7 Future Prospects

Previous works on PCTI have generated promising results that will undoubtedly motivate further research on this topic. Wu et al. (2015) embedded the skeleton of a crevalle jack fish in beeswax and imaged it using a high energy (120 kVp) in-line PCTI system. They showed that detailed structures in the vertebral column of the fish can be clearly distinguished in PCTI, but cannot be easily seen in conventional DBT. The CNR of the spine...
was improved by a factor of more than 2 by using PCTI, and it was further improved by performing phase retrieval prior to tomosynthesis image reconstruction. They also imaged a 6 cm-thick chicken breast with nylon fibril inserts and showed that PCTI with phase retrieval increased the number of distinguishable masses, but also increased the image contrast compared with conventional DBT. Using a grating-based system with a rotating anode medical tube, Li et al. (2014) demonstrated that PCTI can provide improved SDNR and object detectability for several materials. Some of the results are included in this chapter in Figures 52.4, 52.6, and 52.12. Using a synchrotron and grating-based system, Schleede et al. (2014) imaged a formalin-fixed mastectomy specimen from a patient with invasive ductal cancer. They showed that dermal fibrosis, parenchymal necrosis, and cancerous breast tissue can be better differentiated in PCTI than in ACTI.

Note that most published results suggest that PCTI and ACTI provide complementary information to each other, and would be most valuable if used in tandem (Li et al. 2014). In other words, PCTI is likely to become an add-on to conventional ACTI, instead of a competitor. Therefore, one of the important questions to be addressed in future research is how to maintain the image quality and radiation dose efficiency of ACTI, while maximizing the imaging performance potential benefits of PCTI. To answer this question, a thorough understanding on the mutual dependence between PCTI and ACTI will be helpful. For example, the quantitative NPS relationship shown in Equation 52.15 can be used to guide the optimal design of a PCTI system based on an existing ACTI system (Li et al. 2013a), and studies on the anti-scatter effect of the gratings can be utilized to maintain imaging performance of ACTI when designing a PCTI system (Ge et al. 2015).

For clinical breast imaging purposes, the requirement on maintaining the ACTI imaging performance also implies that it is preferable to design a PCTI system based on the geometry and hardware components of a typical DBT system. This is a practical approach towards maintaining the image quality and radiation dose efficiency of ACTI (namely DBT in the context of breast imaging), and it will allow for much easier translation of the PCTI method into clinical environments. Figure 52.19 shows a potential design of a PCTI system based on a clinical DBT system. This is a practical approach towards maintaining the image quality and radiation dose efficiency of ACTI, while maximizing the imaging performance potential benefits of PCTI. To answer this question, a thorough understanding on the mutual dependence between PCTI and ACTI will be helpful. For example, the quantitative NPS relationship shown in Equation 52.15 can be used to guide the optimal design of a PCTI system based on an existing ACTI system (Li et al. 2013a), and studies on the anti-scatter effect of the gratings can be utilized to maintain imaging performance of ACTI when designing a PCTI system (Ge et al. 2015).

For clinical breast imaging purposes, the requirement on maintaining the ACTI imaging performance also implies that it is preferable to design a PCTI system based on the geometry and hardware components of a typical DBT system. This is a practical approach towards maintaining the image quality and radiation dose efficiency of ACTI, while maximizing the imaging performance potential benefits of PCTI. To answer this question, a thorough understanding on the mutual dependence between PCTI and ACTI will be helpful. For example, the quantitative NPS relationship shown in Equation 52.15 can be used to guide the optimal design of a PCTI system based on an existing ACTI system (Li et al. 2013a), and studies on the anti-scatter effect of the gratings can be utilized to maintain imaging performance of ACTI when designing a PCTI system (Ge et al. 2015).

Another important future research topic is how to reduce the image acquisition time of PCTI. For non-synchrotron-based in-line systems, the use of a microfocus X-ray tube will significantly increase the image acquisition time compared with conventional DBT, because the output power of the microfocus tube is much lower compared with the mammography tube. For grating-based PCTI, its imaging time is usually not limited by the tube power, since the method is fully compatible with a diagnostic X-ray tube. On the other hand, the imaging time of PCTI could be severely prolonged, due to the use of a so-called phase stepping procedure (Momose et al. 2003; Weitkamp et al. 2005) in grating-based systems to help extract phase shift information from X-ray intensity measurements. Phase stepping involves multiple mechanical translations of the grating at each view angle; therefore prohibiting continuous gantry rotation, as used in current Hologic DBT systems. To bypass phase stepping and reduce the image acquisition time of grating-based PCTI, a staggered grating design can be used (Ge et al. 2014). Since this method eliminates the translation of gratings during image acquisition, it can also improve the mechanical stability of the PCTI system. Similarly, several other fast phase-contrast image acquisition methods, such as moiré metrology (Bevins et al. 2012) and interlaced phase stepping (Zanette et al. 2011) can be potentially utilized in PCTI to reduce its image acquisition time.

Images shown in this chapter also show that there is still substantial room for further improving the image quality and reducing the image artifacts of PCTI. Although some image artifacts such as out-of-plane signal leakage are intrinsic to tomosynthesis imaging, they can be potentially mitigated by artifact-reduction reconstruction and postprocessing algorithms. Towards this direction, novel and dedicated phase-contrast tomosynthesis reconstruction algorithms need to be developed, in addition to the SAA and FBP methods described in Section 52.4. For example, artifact correction methods originally developed for ACTI (Lauritsch and Harer 1998; Suryanarayanan et al. 2000; Li et al. 2007; Zhang et al. 2008, 2009b) and phase-contrast CT (Gureyev et al. 2007; Lauzier et al. 2010, 2012) can be potentially adopted and customized in PCTI to reduce artifacts and improve image quality.

Although not covered in this chapter, the X-ray dark field (a.k.a. small angle scattering or SAS) signal is another byproduct of grating-based phase-contrast imaging. Results in Schleede et al. (2014) show that dark field tomosynthesis imaging could provide complementary information to PCTI and ACTI. Figure 52.20 shows tomosynthesis images of a cadaver breast specimen with three endogenous contrast mechanisms (absorption, phase, and dark field). All three images were acquired from the same tomosynthesis acquisition using the system described in Li et al. (2014). Blood vessels can be clearly visualized in both dark field and phase images, even in dense regions of the breast. By clearly identifying vessels, some suspicious morphologies of calcifications may be safely dismissed if it is clear that they pertain to a calcified vessel rather than a duct, as in some cancers. Recent
works also suggest that the addition of dark field imaging may allow the composition of microcalcifications in the breast to be determined non-invasively, preventing some biopsies in women with benign calcifications. Therefore, dark field tomosynthesis is another research topic that deserves further study in the future.

As mentioned earlier in the chapter, all PCTI systems reported so far used the inverse Grossman geometry, in which positions of the tube and detectors were fixed. To fully test the feasibility and potential utility of PCTI in clinical imaging, systems with rotating gantry and compact geometry need to be developed, and they should be operated at a clinically acceptable radiation dose level to image thick, fresh, and clinically relevant tissue specimens.

In summary, X-ray phase-contrast tomosynthesis imaging utilizes the wave nature of X-rays to supplement conventional X-ray absorption contrast tomosynthesis imaging, and it reduces the superposition of phase shift signal compared with planar phase-contrast imaging. It can be implemented using several methods, and it is compatible with a diagnostic X-ray tube and conventional detector. The reconstruction methods of phase-contrast tomosynthesis have no fundamental difference with those of absorption contrast tomosynthesis, except the reconstruction kernel. There are plenty of potential opportunities in further improving this imaging method, and its potential applications in preclinical and clinical imaging need to be further explored in the future.

REFERENCES


