Three-dimensional reciprocal space x-ray coherent scattering tomography of two-dimensional object

Zheyuan Zhu and Shuo Pang

The College of Optics and Photonics, CREOL, University of Central Florida, Orlando, FL 32816, USA

(Received 4 June 2017; revised 1 December 2017; accepted for publication 5 February 2018; published 13 March 2018)

Purpose: X-ray coherent scattering tomography is a powerful tool in discriminating biological tissues and bio-compatible materials. Conventional x-ray scattering tomography framework can only resolve isotropic scattering profile under the assumption that the material is amorphous or in powder form, which is not true especially for biological samples with orientation-dependent structure. Previous tomography schemes based on x-ray coherent scattering failed to preserve the scattering pattern from samples with preferred orientations, or required elaborated data acquisition scheme, which could limit its application in practical settings. Here, we demonstrate a simple imaging modality to preserve the anisotropic scattering signal in three-dimensional reciprocal (momentum transfer) space of a two-dimensional sample layer.

Methods: By incorporating detector movement along the direction of x-ray beam, combined with a tomographic data acquisition scheme, we match the five dimensions of the measurements with the five dimensions (three in momentum transfer domain, and two in spatial domain) of the object. We employed a collimated pencil beam of a table-top copper-anode x-ray tube, along with a panel detector to investigate the feasibility of our method.

Results: We have demonstrated x-ray coherent scattering tomographic imaging at a spatial resolution ~2 mm and momentum transfer resolution 0.01 Å⁻¹ for the rotation-invariant scattering direction. For any arbitrary, non-rotation-invariant direction, the same spatial and momentum transfer resolution can be achieved based on the spatial information from the rotation-invariant direction. The reconstructed scattering profile of each pixel from the experiment is consistent with the x-ray diffraction profile of each material. The three-dimensional scattering pattern recovered from the measurement reveals the partially ordered molecular structure of Teflon wrap in our sample.

Conclusions: We extend the applicability of conventional x-ray coherent scattering tomography to the reconstruction of two-dimensional samples with anisotropic scattering profile by introducing additional degree of freedom on the detector. The presented method has the potential to achieve low-cost, high-specificity material discrimination based on x-ray coherent scattering. © 2018 American Association of Physicists in Medicine [https://doi.org/10.1002/mp.12813]

Key words: coherent scattering, computed tomography (CT), small-angle x-ray scattering (SAXS)

1. INTRODUCTION

Biological soft tissues or bio-compatible materials, due to their low atomic weight and similar densities, do not exhibit strong contrast on conventional attenuation-based computed tomography (CT) image. By measuring the elastically scattered x-ray photons from the samples, small-angle x-ray scattering (SAXS) reveals richer information about their molecular composition and structure. X-ray scattering signature has been demonstrated as a high-contrast modality in distinguishing different types of tissues for medical imaging.1–4 If the materials are amorphous or in the fine powder form, the SAXS profile is isotropic, and the tomographic reconstruction of a scalar SAXS profile from each pixel can be obtained from a projection setup similar to conventional CT. This method is termed coherent scattering computed tomography (CSCT), which has shown its success in resolving material-specific scattering profiles,5–7 and has also been demonstrated feasible in medical imaging settings.8,9 The requirement on isotropic scattering signal guarantees that the scattering from an object voxel remains unchanged regardless of the incident direction of x-ray, and thus, the projection data contributed from the same voxel is invariant under different projection angles. As a result, conventional CSCT can only reconstruct the coherent scattering profile in terms of the magnitude of momentum transfer, not the complete SAXS pattern, which is a function of the three-dimensional momentum transfer vector. However, the isotropic scattering assumption on the material is not true in some biological or bio-compatible samples. For example, tissues such as bones and teeth,10,11 as well as bio-compatible polymers used in medical implants,12 possess anisotropic scattering profile due to the existence of partially ordered structure at molecular level. Imaging the cellular formation within biomaterials is an indispensable step in evaluating the different
2. METHODS AND MATERIALS

2.A. Direction-dependent x-ray coherent scattering

We start with resolving the scattering pattern associated with a row of pixels integrated along a single x-ray beam. Figure 1 shows the geometry of a pencil-beam x-ray coherent scattering setup. The system employs a copper-anode x-ray tube (XRT60, Proto Manufacturing), which emits a sharp Kz line ($\Delta E / E_0 < 0.1$) centering around $E_0 = 8$ keV, as shown in Fig. 2. The incident x-ray spectrum can be approximated by a delta-function $S(E) = I_0 \delta(E - E_0)$. The diverging x-ray source is collimated by a pair of pinholes to a narrow beam propagating along the $z$ direction. The number of coherent scattering photons $dl$ from a single object voxel toward the detector pixel located at $r_d$ is given by the following:

$$dl = I_0 \int_{E} S(E) \frac{r_d^2}{2} (1 + \cos^2 \theta) \varphi(t, q) \beta(t, \theta) d\Omega dV d\mathbf{q} dE$$

(1)

where $r_\text{e}$ is the classical electron radius, $\theta$ is the scattering angle, and the $\cos \theta \approx 1$ for small-angle scattering. The object function $f(t, q) = n_0(t)/F(t, q)$, which is the product between material density $n_0(t)$ and the square of molecular form factor $F(t, q)$; $t$ is the depth of object voxel along the beam; $dV = d\mathbf{r}d\mathbf{z}$, which is the volume of the object voxel illuminated by the x-ray beam with cross-sectional area $A$; $d\Omega = \Delta^2 / |r_d - \mathbf{t}|^2$, which denotes the solid angle covered by a detector pixel with pixel size $\Delta$ located at $r_d = (x_d, y_d, z_d)$. The delta-function spectrum integrates to a constant $I_0$. $\varphi(t)$ and $\beta(t, \theta)$ represent the attenuation on the incident and scattered beams, respectively. The momentum transfer vector $\mathbf{q} = \mathbf{k}' - \mathbf{k}_0$, defined as the change in the wave vector between incident and scattered x-ray, is given by the vector form of Bragg’s law:

$$\mathbf{q} = \frac{E}{hc} \left( \frac{r_d - \mathbf{t}_z}{|r_d - \mathbf{t}_z|} - \hat{z} \right),$$

(2)

where $E$ denotes the energy of the quasi-monochromatic x-ray source, $h$ is Planck constant, and $c$ is the speed of light.

![Figure 1. Pencil-beam coherent scattering model applied to vector momentum transfer.](image-url)
According to Eq. (2), the detector plane at a fixed location, \( z_d \), covers a curved Ewald sphere in three-dimensional \( q \) space. If the distance between the object voxel and detector, \( (z_d - t) \), is much greater than the horizontal (vertical) offset of the detector pixel, \( x_d(y_d) \), we can ignore them on the denominator. The vector form of Bragg’s law can be simplified to

\[
q = \frac{E}{hc} \left( \frac{x_d \hat{x} + y_d \hat{y}}{z_d - t} \right),
\]

which indicates that under small-angle approximation, the Ewald sphere becomes a flat section defined by the plane \( q_z = 0 \) in the momentum transfer space. The detector plane can be expressed in terms of the cylindrical coordinates by defining \( w = (x_d^2 + y_d^2)^{1/2} \) and \( \psi = \arctan(y_d/x_d) \) as the distance and azimuthal angle of each detector pixel with respect to the center of detector. Rewriting Eq. (3) in cylindrical coordinates and combining with Eq. (1), the intensity detected by the detector pixel located at \( r_d \) can be expressed in the cylindrical coordinate as follows:

\[
g(w, z_d, \psi) = \int r_d^2 f(t, q) \delta \left[ q - \frac{E}{hc} \left( \frac{w \hat{r}}{z_d - t} \right) \right] \frac{\Delta^2}{w^2 + (z_d - t)^2} \mathrm{d}qd\psi,
\]

where \( \hat{r} = (\cos \psi, \sin \psi, 0) \) is the unit vector along the radial direction. Due to the confinement of the scattering volume in the x-y plane under pencil-beam illumination, the q vector probed by Bragg’s law contains only the radial component \( q = Ew/(hc(z_d - t)) \hat{r} \). Therefore, the radial and azimuthal directions in the momentum transfer space can be separated by expressing the q vector on the section of \( q_z = 0 \) plane as \( q = q_r \hat{r} \), where \( q_r = (q_x^2 + q_y^2)^{1/2} \) is the magnitude of the q vector on \( q_z = 0 \) plane and \( \hat{r} \) is the unit vector along radial direction under azimuthal angle \( \psi = \arctan(q_y/q_x) \). As a result, the integral on the \( q - t \) hyperplane in Eq. (4) is confined to the 2D \( q_z - t \) plane under each azimuthal angle \( \psi \). By moving the detector and capturing different scattering images along \( z \) direction, the resulting measurement establishes a linear mapping between the object domain \( (t, q_z) \) and the measurement domain \( (w, z_d) \). By discretizing the object with sampling rate \( \Delta q \) in spatial domain and \( \Delta q_z \) in momentum transfer domain, and partitioning the panel detector into square pixels with pitch \( \Delta q \) and step size \( \Delta q_z \) along the beam direction, Eq. (4) can be written in the form of a linear system \( g = Hf \), where the system matrix is constructed based on the weight of each object voxel calculated from the term \( \Delta^2/(w^2 + (z_d - t)^2) \mathrm{d}rdq \). A special case of this model using the scattering images from two detector locations \( z_d \) has been demonstrated.\(^{21}\)

### 2.B. Measurement scheme

The three degrees of freedom in the pencil-beam measurement, two from the panel detector and one from the detector movement, enable the recovery of three-dimensional scattering volume along each pixel hit by x-ray. To resolve the three-dimensional \( q \)-space within a two-dimensional sample slice, a tomographic data acquisition scheme, combining the sample rotation \( \phi \) around \( y \) direction and translation \( s \) along \( x \) direction, is introduced to the pencil-beam system model, as shown in Fig. 3(a). To facilitate the discussion of the tomography geometry, we establish an object coordinate that translates and rotates with the sample, and express a total of five dimensions associated with the spatial and momentum transfer domain of the sample in terms of the object coordinate \( f_o(r_o, q_o) \). Without any rotation or sample translation, the \( x, y, \) and \( z \) directions of the sample coordinate co-align with those of the laboratory coordinate. The slice imaged by our system is defined as \( y_o = 0 \). Under specific projection angle, \( \phi \), and sample translation, \( s \), the object function probed by the pencil-beam coherent scattering system becomes a series of 2D sections in momentum transfer space along a row of voxels illuminated by the beam:

\[
f_{s, \phi}(t, q) = f_o(A_{\phi}(s, 0, t)^T, A_{\phi}q),
\]

where \( A_{\phi} \) is the 3 by 3 rotation matrix around \( y \) axis with rotation angle \( \phi \). Combining the pencil-beam model in Eq. (4) with the tomographic geometry in Eq. (5), a total of five dimensions are collected in our measurement scheme for the reconstruction of \( f_o(x, y, z, s, q_x, q_y, q_z) \):

\[
g_{s, \phi}(w, z_d, \psi) = \int r_d^2 f_{s, \phi}(A_{\phi}(s, 0, t)^T, A_{\phi}q) \delta \left[ q - \frac{E}{hc} \left( \frac{w \hat{r}}{z_d - t} \right) \right] \frac{\Delta^2}{w^2 + (z_d - t)^2} \mathrm{d}qd\psi.
\]

Our x-ray tube (XRT60, Prototyping Manufacturing) is a copper-anode source operating under 45 kV and 40 mA without any filtration. The spectrum of the source is measured with a
which establishes a backprojection from the measurement data to an extended object, which consists of sample translation and rotation in addition to the pencil-beam coherent scattering model. (b) The five-dimensional measurement consists of a series of 4D sinograms along the z direction. Each pixel in the sinogram under a specific rotation in addition to the pencil-beam coherent scattering model. (b) The five-dimensional measurement as a result of sample shift, rotation, and detector translation is shown in Fig. 3(b), which is a series of four-dimensional sinograms along z. Each pixel on the 4D sinogram contains an entire 2D scattering image $g(x_d, y_d)$ rather than a single value as the attenuation-based CT.

### 3. RECONSTRUCTION AND RESULTS

For a fixed detector location, $z_d$, and $\psi = 90^\circ$ on the detector plane, the unit vector $\hat{r}$ in Eq. (4) becomes $\hat{y}$, and therefore, only the momentum transfer vector $q_y$ is captured by the pencil-beam system. Since the vector $\hat{y}$ is invariant to matrix $A_\phi$ under all projection angles, the same momentum transfer $q_y$ is probed for all the object voxels as the sample rotates, which matches the assumption of conventional CSCT. Eq. (6) can be simplified for $\psi = 90^\circ$ as

$$
g_{s,\phi}(w) = r_e^2 \frac{\Delta^2}{w^2 + (z_d - t)^2} A$$

$$
\int f(s \cos \phi + t \sin \phi, 0, -s \sin \phi + t \cos \phi, \frac{w}{\hbar c} \left( \frac{w}{z_d - t} \right)^2 \psi) dt.
$$

The projection transform in Eq. (7) can be treated as a line integral on the $q_y - t$ plane along a family of curves $q_y = Ew/(\hbar c(z_d - t))$ with $w$ as a parameter. The tomographic reconstruction on a subset of the object function $f_o(x, z, 0, q_y, 0)$ is implemented using the angular-dispersive coherent scattering computed tomography (AD-CSCT) scheme, which establishes a backprojection from the measurement domain $(s, \phi, w)$ to the object domain $(x, z, q_y)$. The sampling is 1 mm in the spatial domain, and 0.005 Å$^{-1}$ along $q_y$ direction. The AD-CSCT reconstruction from a subset of measurement data $g_{s,\phi}(w; z_d = 120mm, \psi = 90^\circ)$ is summarized in Fig. 4.

Figure 4(a) shows a normalized transmission map from a CT scan of the sample. The spatial profiles of the sample reconstructed from the intensity along the vertical radial direction on the detector are shown in Fig. 4(b1)–4(b3), corresponding to $q_y = 0.08, 0.12$ and 0.16 Å$^{-1}$, respectively. On a conventional CT image, water appears stronger than butter due to its high x-ray attenuation coefficient. However, the contrast between water and butter exhibits differently on CSCT images due to the higher scattering peak of water compared with butter. Therefore, at low momentum transfer, water appears to be a weaker scatter than butter, while the contrast between water and the butter reverses at high momentum transfer value. The reconstructed scattering
profiles of three pixels marked in green, blue, and red, corresponding to butter, water and Teflon, are compared with their reference profiles in Fig. 4(c1)–4(c3), respectively. To collect the reference scattering profile, each material was shaped into a 2-mm-wide tip, illuminated by the same pencil beam used for tomography scan. Only the intensity along the vertical direction on the resultant 2D scattering image was picked as the reference. Figure 4(c) indicates a high consistency between the reconstructed scattering profile and that of the individual material, with error mainly in the high momentum transfer values, under which the scattering signal-to-noise ratio is low.

Both the setup geometry and the energy spectrum of the source contribute to the momentum transfer resolution of conventional CSCT. Differentiating the Bragg’s law along the radial direction in Eq. (4) with respect to \( t \), \( w \), and \( E \) yields the uncertainty of the momentum transfer, \( \Delta q \)

\[
\Delta q = \sqrt{\left(\frac{E \Delta w}{hc z_d - t}\right)^2 + \left(\frac{E w \Delta t}{hc (z_d - t)}\right)^2 + \left(\frac{\Delta E}{hc (z_d - t)}\right)^2}, \tag{8}
\]

where \( \Delta w = 0.4 \) mm is the effective pixel size of the detector after binning; \( \Delta E \) is the spectrum width at the K\( \alpha \) peak (\( \Delta E/E_0 = 0.09 \)). For the AD-CSCT reconstruction along the vertical scattering direction under \( z_d = 120 \) mm, the spatial resolution perpendicular and parallel to the beam direction are both determined by the beam spot size \( \Delta t = \Delta s = 2 \) mm.

Considering a scattering angle of 10°, which is roughly the peak position of water in our setup, located at lateral position \( t = 0 \), the uncertainty in the momentum transfer domain is about \( \Delta q_y = 0.01 \) Å\(^{-1}\). Notice that for pixels closer to the detector, the momentum transfer uncertainty will become larger and vice versa. We sample the momentum transfer along the \( y \) direction at the Nyquist interval 0.005 Å\(^{-1}\) in the AD-CSCT reconstruction. Note that the beam hardening of the incident and scattered radiation will further blur the reconstructed scattering profile.

Except for the rotation-invariant radial direction \( \psi = 90° \) on the detector, all other radial directions cannot be directly mapped back to the \( q_r \) under the corresponding azimuthal angle using conventional AD-CSCT reconstruction, since different 2D momentum transfer sections are probed as projection angle changes. Instead, recovering the scattering profile for all voxels along the beam \( f(t, q_r, \psi) \) relies on the inversion of the pencil-beam model described by Eq. (4). However, due to the small scattering angle and limited range of detector movement, direct inverse of the pencil-beam system model in Eq.(4) would result in voxels heavily blurred along the beam, a phenomenon that also appears in Ref. [21]. To decouple the scattering direction from different positions along the beam, prior information on the spatial distribution of the sample obtained from AD-CSCT was used to assist the reconstruction. First, the support of the object was acquired by thresholding the AD-CSCT reconstruction in Fig. 4 (The background is less than 10% of the maximum intensity). We
could then obtain a list of nonzero pixels along each pencil beam by rotating the support to different projection angles. For each pencil beam, all the scattering data under a specific azimuthal angle $\psi$ was extracted from the measurement and stored as $g(w, z_\theta)$. Then, based on the discretized forward model matrix $H$ from Eq. (4), $f(t, q_r)$ at every azimuthal angle was reconstructed with the object support enforced along dimension $t$ by solving the optimization problem:\cite{equation}

$$\hat{f} = \arg\min_{f} \left(- \log P(\mathbf{g} | \mathbf{f}) + \tau \sum_{i \in \text{support}} f_i \right), \quad (9)$$

where $P(\mathbf{g} | \mathbf{f})$ is the Poisson likelihood of observing the measurement $\mathbf{g}$ given the parameters $\mathbf{f}$; $\tau$ is the weight for balancing the measurement error and the regularizer, which penalizes the results with large values outside the support. The reconstructions under $2\pi$ azimuthal angle $\psi$ were grouped together to form one section of the 3D momentum transfer space $f(t, q_r, \psi)$ along each pencil beam. The five dimensions in sample coordinates were recovered by combining all the pencil-beam positions $s$ and rotation angle $\phi$.

Figure 5(a) displays the spatial profile under $q_z = 0.08 \, \text{Å}^{-1}$, which is reconstructed from the $\psi = 0^\circ$ direction on all the scattering images under rotation $\phi = 0^\circ$. Teflon wrap exhibits strong scattering mainly in the $y$ direction due to the formation of molecular chains aligned predominantly along the horizontal direction, where the polymer film is stretched. As a result, Teflon is visible in the spatial profile reconstructed from $y$ direction in Fig. 4(b), but is hardly visible in the spatial profile in Fig. 5(a). The three-dimensional scattering profiles of the pixel marked in green, blue, and red, corresponding to the material of butter, Teflon, and water, are shown in Fig. 5(b1)–5(b3), respectively. The display window of the scattering profile covers $|q|$ from 0 to 0.3 Å$^{-1}$. Since butter and water are both noncrystalline material, their scattering intensities depend only on the magnitude of the momentum transfer vector $|q|$, which agrees with the ring-shaped scattering profile reconstructed from our experiment. Seven sections of the three-dimensional $q$-space of Teflon wrap ranging from $0^\circ$ to $180^\circ$ projection at an angular step of $30^\circ$ are shown in Fig. 5(c). By enclosing all pixels inside the contour of 70% maximum scattering intensity with minimal bounding circles,\cite{equation} the center of two scattering peaks can be extracted, and then, the tilt angle of these two peaks with respect to $y$ axis can, thus, be calculated. The tilt angles of the two strongest scattering spots in the scattering profile against vertical direction are found to be $4.4^\circ$, $3.4^\circ$, $3.0^\circ$, $1.0^\circ$, $2.0^\circ$, $3.1^\circ$, and $4.4^\circ$, respectively. The decreasing trend is consistent with the orientation of Teflon wrap projected onto the x-y plane under these rotation angles.

4. DISCUSSION

4. A. Spatial and momentum transfer resolution along non-rotation-invariant directions

It is worthwhile to notice that compared with CSCT, sample rotation plays a different role in our system. In conventional CSCT, rotation assists in resolving the spatial distribution of the sample since the scattering profile of a specific sample voxel is assumed to be consistent along different projection angles. For materials with anisotropic scattering profile, this condition holds only for the scattering intensity along a rotation-invariant direction parallel to the rotation axis, and therefore, only the tomography reconstruction from the $\psi = 90^\circ$ direction on the detector yields the correct spatial profile. For the remaining majority of momentum transfer vectors, separating the scattering profile corresponding to different pixels primarily depends on the inversion of pencil-beam system model, while the function of rotation is limited to probing different sections in the three-dimensional momentum transfer volume.

Here, we would like to further discuss the achievable spatial and momentum transfer resolution from the pencil-beam system model with detector scan along the beam.

![Fig. 5. (a) The spatial profile of the sample under $q_z = 0.08 \, \text{Å}^{-1}$. (b1–b3) Two orthogonal sections in the three-dimensional scattering volume of the pixel in the region of butter, water, and Teflon wrap, respectively. (c) The scattering volume of Teflon is expanded into seven slices under projection angles of $0^\circ$ to $180^\circ$ at $30^\circ$ step. The white, dashed line indicates the $y$ direction to facilitate the display of the orientation of scattering peak as the sample undergoes rotation. The display window of the momentum transfer space is 0–0.3 Å$^{-1}$. [Color figure can be viewed at wileyonlinelibrary.com]](image-url)
direction. If we pick a specific radial direction on the detector, each detector pixel located at \( w \) from the center of detector captures an integral in the \( q_r - t \) domain along a curve defined by Eq. (4). At specific detector position, \( z_d \), all detector pixels along the radial direction constitute a family of curves on the \( q_r - t \) plane, with the gradient of the curves determined by \( \Delta z_d \). Figure 6(a) demonstrates a series of curves separated by \( \Delta w \) under the smallest and largest \( z_d \), shown in red and blue, respectively. The gradient of these curves increases as the detector is moved closer to the sample. For a specific \( z_d \), the radial intensity distribution \( g(w) \) is equivalent to a projection of the 2D object function \( f(t, q_r) \) under an angle determined by the gradient. The allowable range of detector movement along \( z \) direction confines the gradient of these curves, thus restricting the projection angles in the \( q_r - t \) domain, which is analogous to the limited-angle CT problem. The location of the voxel and its scattering angle are highly coupled in the \( q_r - t \) domain, which cannot be fully decoupled with limited projection angle. The green intersection area in Fig. 6(a) illustrates the achievable resolution \( \Delta r \approx 28 \text{ mm} \) and \( \Delta q_r \approx 0.015 \text{ Å}\(^{-1}\) without any prior information. Figure 6(b1) shows a reconstruction along a single beam penetrating through the Teflon and water under \( \phi = 0^\circ \), \( \psi = 90^\circ \) by directly inverting the pencil-beam system mode in Eq. (4). Due to the ambiguity in the \( q_r - t \) plane, two materials cannot be distinguished along the beam. If the spatial distribution of these two materials are reconstructed from the AD-CSCT measurement and then enforced as the spatial support in the reconstruction of other momentum transfer vectors, the reconstruction, shown in Fig. 6(b2), can distinguish these two materials with the same resolution in reciprocal space as the AD-CSCT. Notice that the spatial and momentum transfer resolution of the pencil-beam system can be further improved using larger detector, thus allowing the detector to move over a wider \( z_d \) range without missing the high momentum transfer information.

4.B. Attenuation and beam hardening

In our experiment, we did not correct for the attenuation on the scattered beam due to the relatively small size and simple geometry of the sample. For large objects, the attenuation needs to be calibrated and included in the reconstruction. Eq. (1) contains two mechanisms of attenuation: the reduction of incident photon flux by the voxels upstream the pencil-beam \( x(t) \) and the attenuation of scattered photons along the scattering direction \( \beta(t, \theta) \). If the scattering angle is small, the scattered beam follows almost the same path as the transmitted beam. Therefore, the self-attenuation can be corrected by normalizing the scattered pattern with transmitted photon flux. When the scattering angle is large, especially under low energies, three-dimensional attenuation map is needed to compensate for the attenuation of the scattered beam.

The attenuation on the low-energy photons by the sample can also cause beam hardening, which shifts the spectrum of the scattered beam to a higher energy than incident spectrum. Our reconstruction was performed under Kz line of copper anode \( E_0 \approx 8 \text{ keV} \). With the presence of beam-hardening effect, the reconstructed object function in Eq. (1) is distorted by the spectrum \( f'(q) = \int S(E)f(q \cdot E/E_0)dE \), where \( f(q) \) is directly proportional to the square of molecular form factor. This distortion could cause the shift of the scattering peak to lower momentum transfer, and the loss of fine features in the reconstructed scattering profile. To quantify the beam-hardening effect, 2D energy-sensitive detector array is needed to directly measure the spectrum of scattered photons. The beam hardening can be reduced by switching to a high-energy source with proper filtration, such as molybdenum or silver-anode x-ray tubes.

5. CONCLUSIONS

In summary, we have developed a method to probe the direction-dependent x-ray scattering information inside an extended object. By introducing detector scan along the beam and tomographic data acquisition technique into a
pencil-beam coherent scattering model, we are able to reconstruct the three-dimensional momentum transfer space of a two-dimensional sample slice. Compared with existing SAXS tomography that preserves anisotropic scattering profile, our proposed setup uses only a table-top x-ray tube and eliminates the tilt of rotation axis, making it easily accessible for biomedical applications. In addition, our method decouples the spatial and momentum transfer space for the non-rotation-invariant directions on the detector. As a result, the number of projections, under which a detector scan along z direction is performed, can be picked based on the number of sections required in 3D q-space. However, given the size of our detector, there is a \( \sim 15^\circ \) missing cone in the 3D q-space that is not covered by our measurement without tilting the sample out of the rotation plane. This imaging modality may find potential biomedical applications including monitoring the tissue regeneration in porous media and assessing the bone

Partially textured 3D reciprocal-space SAXS profile could also be used as a contrast mechanism in security screening applications. The 3D reciprocal-space coherent scattering tomography could serve as a stand-alone, high-contrast imaging modality in medical imaging, or a secondary screening method in addition to conventional CSCT in case anisotropic scattering signal is present.

**ACKNOWLEDGMENTS**

The research is supported by the National Science Foundation (DMS-1615124) and Oak Ridge Associated Universities Ralph Powe Junior Faculty Award. The authors have no conflicts to disclose.

**REFERENCES**

ClearCheck
One-click plan evaluation

Dose Constraints • Plan Checks
Structure Checks • Collision Checks

RADformation.com