Development of x-ray coherent scatter projection imaging systems

by

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Abstract

Coherent scatter x-ray imaging systems are sensitive to material structure and chemical composition, and generate images with contrast superior to conventional transmission x-ray imaging. The goal of this thesis was to develop a scatter projection imaging system capable of acquiring images in less than ten minutes, in the range of nuclear medicine scans, to be practical for future applications such as bone specimen imaging.

Two systems were developed. The first improved on the group's previous work and was configured at the Canadian Light Source synchrotron. It employed five 33.2 keV pencil beams in combination with continuous object motion. The system consisted of a primary collimator, motorized stages for object translation, a flat-panel x-ray detector for measuring scattered x rays, and discrete photodiodes for simultaneously measuring transmitted x rays. The acquisition time for a $5.0\text{ cm} \times 9.0\text{ cm}$ object with 8425 pixels was 2.3 min. Use of continuous motion acquisition increases the width of image boundaries by the product of the object translation speed and the acquisition time per pixel. Contrast-detail performance was independent of acquisition speed. Pixel signal-to-noise ratio (SNR) measurements indicated that the scatter data were limited by the detector readout noise.

The synchrotron system was a development stage in an idealized environment but a practical system requires a commonly available x-ray source. Therefore, a second system was developed using a conventional rotating-anode x-ray tube. An array of up to three rows by five columns of pencil beams can irradiate the object simultaneously. A 110 kVp spectrum with 2.25 mm of added Al filtration was used. Motorized stages translate the object through the beams for step-and-shoot acquisition. For this first x-ray tube-based system, the primary imaging capability was not optimized and transmitted x rays are mea-

sured through attenuators using the same flat-panel detector that measures scattered x rays. The shortest acquisition time for a $6.0\text{ cm} \times 10.0\text{ cm}$ object with 6000 pixels was 8.8 min. Contrast-detail performance was similar to the synchrotron-based system. SNR measurements indicated that the scatter data were not readout noise limited.

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List of Symbols and Abbreviations

$\mathbf{A}$	system matrix linking scatter profile intensities and detector element values
A_1	x-ray exposure level at a detector element prior to beam crossing a sharp material boundary
A_2	x-ray exposure level at a detector element following beam crossing a sharp material boundary
A_A	pixel value above a horizontal boundary in an image
A_B	pixel value below a horizontal boundary in an image
A_L	pixel value to the left of a vertical boundary in an image
A_R	pixel value to the right of a vertical boundary in an image
ADC	analog-to-digital converter
a	translation stage acceleration
a_{ij}	element of matrix $\mathbf{A}$
B	magnitude of the magnetic field of a bending magnet
B_0	maximum magnitude of the magnetic field of an insertion device
BMD	bone mineral density
BMIT	Biomedical Imaging and Therapy facility
BMIT-ID	Insertion Device beamline of BMIT
C	subject contrast
C_P	subject contrast for primary transmission imaging
C_S	subject contrast for scatter imaging
CLS	Canadian Light Source synchrotron
CNN	convolutional neural network
CT	computed tomography
c	speed of light

$D_{m,n}$	offset of the element in row m and column n of a flat-panel detector
DT	direct tomography
DXA	dual energy x-ray absorptiometry
d	object midplane-to-detector distance in x-ray tube-based system
Δd	path length difference between two scattering trajectories
E	energy of a photon
E_b	binding energy of an electron
$\mathcal{E}_e$	total energy of an electron
$E(x)$	horizontal profile of a real vertical boundary
$E(y)$	vertical profile of a real horizontal boundary
$E_0(x)$	horizontal profile of an ideal vertical boundary
$E_0(x, y)$	ideal horizontal boundary
e	magnitude of the electron charge
e	base of the natural logarithm
$F_{coh}(\vec{x}, \rho_e(\vec{r}))$	coherent scatter form factor
$eF_{coh}(\vec{x}, \rho_e(\vec{r}))$	coherent scatter form factor per electron
$F_{KN}(h\nu, \theta)$	Klein-Nishina factor
F^θ	factor that scales the exposure incident on an object to the scatter exposure incident on a flat-panel detector element
$\vec{f}$	vector of scatter profile intensities for a scatter pattern
f_j	element of vector $\vec{f}$
$f(k)$	recorded output of a flat-panel detector including lag for acquisition k
GUI	graphical user interface
$\vec{g}$	vector of detector element values for a scatter pattern
g_i	element of vector $\vec{g}$
$g(k)$	ideal output of a flat-panel detector if there was no lag for acquisition k
$H(x)$	Heaviside unit-step function
HPGe	high-purity germanium
h	Planck's constant
$h(k)$	impulse response function of a flat-panel detector

$h(x, y)$	impulse response for aperture blur of a horizontal boundary using the x-ray tube-based system
$h_{aperture}(x)$	impulse response for the horizontal aperture blur of a vertical boundary in an image
$h_{motion}(x)$	impulse response for motion blur
IRF	impulse response function
i	imaginary unit
K	undulator parameter
K_{air}	object-incident air kerma
kVp	peak kilovoltage of an x-ray generator
LSB	least significant bit
LTI	linear time-invariant
M	number of exponential terms in the IRF of a flat-panel detector
MLEM	maximum likelihood-expectation maximization
mAs	product of the average x-ray tube current, in mA, and the exposure time, in s
m_e	electron mass
N	number of photons having undergone no interaction after traversing some thickness of a material
N_0	number of photons incident on an object
N_{beams}	number of beams used for a scatter pattern
N_{per}	number of periods in an insertion device
n	harmonic number of undulator radiation
P	probability that a photon has undergone at least one interaction after traversing some thickness of a material
PCI	phase-contrast x-ray imaging
$P_{m,n}^{meas}$	measured output of the element in row m and column n of a flat-panel detector
$P_{m,n}^{flood}$	measured output of the element in row m and column n of a flat-panel detector when directly exposed to a flood field
$P_{m,n}^{corr,usual}$	corrected output of the element in row m and column n of a flat-panel detector when using the usual calibration method
PBS	phosphate-buffered saline
p	number of x-ray generator waveform pulses in a 1/60 s interval

$\vec{p}$	momentum of a photon
$\vec{p}'$	momentum of a scattered photon
$\hat{p}$	unit vector parallel to the momentum vector of a photon
$\hat{p}'$	unit vector parallel to the momentum vector of a scattered photon
RF	radio frequency
$\vec{r}$	position vector of an electron
r_0	classical electron radius
r_1	smaller radius of the conical frustum of the pencil beams in the x-ray tube-based system
r_2	larger radius of the conical frustum of the pencil beams in the x-ray tube-based system
S	total scattering amplitude
$S_{inc}(x, Z)$	incoherent scattering function
$S_{m,n}^{true}$	actual sensitivity of the element in row m and column n of a flat-panel detector
$S_{m,n}^{calc,usual}$	sensitivity of the element in row m and column n of a flat-panel detector estimated using the usual calibration method
$\overline{S_{calc,usual}}$	mean of the estimated sensitivities of the elements of a flat-panel detector
SAXS	small-angle x-ray scattering
SDNR	signal-difference-to-noise ratio
SNR	signal-to-noise ratio
s	source-to-object midplane distance in x-ray tube-based system
$\vec{s}$	twice the momentum transfer argument
T	kinetic energy of an electron
TFT	thin-film transistor
t	thickness of a material
t_B	thickness of the PMMA background of the contrast-detail phantom
t_D	thickness of a PMMA disc of the contrast-detail phantom
Δt	pulse train period or pixel acquisition time for continuous motion acquisitions
Δt_C	time interval between repeated x-ray tube exposures

Δt_E	duration of x-ray tube exposure
U	thermal energy stored in x-ray tube anode
U_{0C}	thermal energy stored in x-ray tube anode at the end of an exposure
U_{0E}	thermal energy stored in x-ray tube anode at the start of an exposure
U_{eqI}	equilibrium thermal energy stored in x-ray tube anode if an image scan continued indefinitely
V	volume
v	stage speed for continuous motion acquisitions
W_x	horizontal width of a vertical boundary
W_y	vertical width of a horizontal boundary
X	exposure
$X^{flood}(x, y)$	spatially non-uniform exposure flood field (continuous)
$X_{m,n}^{flood}$	spatially non-uniform exposure flood field (discrete)
$\overline{X^{flood}}$	mean flood field exposure over a flat-panel detector
$\vec{x}$	momentum transfer argument
$x_{c,k}$	corrected horizontal stage motor position for pixel k
$x_{m,k}$	measured horizontal stage motor position for pixel k
x_0	location of a vertical boundary in an image
Δx	horizontal spacing between pixels for continuous motion acquisitions
y_0	location of a horizontal boundary in an image
Z	atomic number
Z_{coh}	number of electrons in the fundamental unit of a substance
α	ratio of a photon's energy to the energy of an electron in its rest frame
$\alpha(x, y)$	exposure variation factor of flood field (continuous)
$\alpha_{m,n}$	exposure variation factor of flood field (discrete)
$\alpha_{m',n'}$	exposure variation factor of the beamlet (pencil beam) used for scatter imaging
α_m	relative strength of the m th exponential term in the IRF of a flat-panel detector
$\beta_1, \beta_2, \beta_3$	parameters of exponential fit to x-ray tube cooling curve data

γ	Lorentz factor
$\hat{\epsilon}$	unit vector parallel to a photon's electric vector
$\hat{\epsilon}'$	unit vector parallel to a scattered photon's electric vector
θ	photon scattering angle
ϑ	anode angle
λ	wavelength of a photon
λ_u	spatial period of an insertion device
μ	linear attenuation coefficient
μ_k	partial linear attenuation coefficient for interaction k
ν	frequency of a photon
ν'	frequency of a scattered photon
ν_1	fundamental frequency of undulator radiation
ν_c	central frequency of bending magnet radiation
π	ratio of a circle's circumference to its diameter
$\rho_e(\vec{r})$	electron density distribution (number of electrons per unit volume)
${}_e\sigma_k$	total scatter cross section per electron for interaction k
$\frac{d_e\sigma}{d\Omega}$	Thomson differential scattering cross section for a polarized beam of x rays
$\frac{d_e\sigma_0}{d\Omega}$	Thomson differential scattering cross section for an unpolarized beam of x rays
$\frac{d\sigma_{coh}}{d\Omega}$	coherent scatter differential cross section
$\frac{d_e\sigma_{coh}}{d\Omega}$	coherent scatter differential cross section per electron
$\frac{d\sigma_{inc}}{d\Omega}$	incoherent scatter differential cross section
$\frac{d_e\sigma_{inc}}{d\Omega}$	incoherent scatter differential cross section per electron
$\frac{d_e\sigma_{KN}}{d\Omega}$	Klein-Nishina differential cross section per electron
$\frac{d_e\sigma_k}{d\Omega}$	differential scatter cross section per electron for interaction k
${}_e\tau$	photoelectric cross section per electron
τ_{FP}	delay between the time that the object's position corresponds to the true scatter image pixel centre and the time that the stage motor positions are sampled
τ_{PD}	delay between the time that the object's position corresponds to the true primary image pixel centre and the time that the stage motor positions are sampled

τ_m	decay constant of the m th exponential term in the IRF of a flat-panel detector
τ_x	width of a rectangular pencil beam
Φ_0	x-ray fluence incident on the contrast-detail phantom
$\Phi(t)$	x-ray fluence after passing through PMMA of thickness t
ϕ	angle between the trajectories of a scattered electron and an incident photon
φ	phase of the probability amplitude of a photon trajectory
Ψ_j	scattering amplitude for a trajectory with index j
ψ	azimuthal angle of photon scattering
ω	angular velocity of a rotating anode

Chapter 1

Introduction

Wilhelm Röntgen¹ discovered x rays on 8 November 1895 while investigating the effects of discharges from electrodes in vacuum tube equipment.² He subsequently studied the properties of these newly discovered rays and their ability to penetrate different materials. In the course of his investigations he initiated the field of radiological imaging, having taken a radiograph of his wife's hand. Within a few weeks, Röntgen's discovery and its implications for medicine spread worldwide.²

The principle of radiography is the casting of shadows by an object when it is placed in the path of a source of x rays, enabling the visualization of its internal structure. Of interest are the transmitted x rays. Different materials in the object attenuate x rays to a greater or lesser degree based on how strongly the x rays interact with them. The probability of interaction per unit length is called the material's linear attenuation coefficient, μ , and it depends on the energy of the x rays. The material properties affecting attenuation are primarily the effective atomic number and the density. More dense, higher atomic number materials, such as bone or metal, are more attenuating than less dense, lower atomic number materials, such as muscle or plastic. Increasing the thickness of a material also increases attenuation. A radiograph is the projection or line integral of the spatially-dependent linear

attenuation coefficient along each ray through the object from a point-like source to some x-ray sensitive detector.

The attenuation of the primary beam is not only due to absorption of the x rays but also due to scattering. That x rays can scatter or “bend” was first demonstrated in 1899 by Dutch physicists Haga and Wind³ who observed the broadening of the image of a slit when illuminated with x rays. In the energy range of x rays used for medical, industrial, and security radiography, there are two scattering processes, coherent (Rayleigh) scattering and incoherent (Compton) scattering.⁴

Scattered x rays are always present when an object is irradiated. In conventional radiography, the transmitted primary x rays are of interest and scatter reduces contrast and signal-to-noise ratio (SNR) in images.⁴ Photons scattered out of one primary ray in the x-ray beam can reach the image receptor at a location where some other primary ray is incident, misrepresenting the transmission along the second ray since scattered and unscattered x rays are usually not distinguished. It is therefore desirable to minimize the amount of scattered radiation reaching the detector by using an anti-scatter grid or by some other method. Figure 1.1 is a diagram of a typical radiography system for medical imaging.

In the decades following the discovery of x rays, many technological developments have improved the diagnostic capability of transmission radiography. Radiation source developments include the use of a heated filament coil as a source of electrons and the introduction of high power rotating anode x-ray tubes.² Image receptors have evolved from the barium platinocyanide fluorescent screens and photographic dry plates used by Röntgen¹ to the digital flat-panel detectors commonly used today. System developments include Hounsfield’s computed tomography (CT) scanner,⁵ which by the acquisition of x-ray projections from a multitude of angles allowed for the reconstruction of cross-sectional image slices of an object that provided soft-tissue contrast much higher than previously attainable.

The effectiveness of x-ray radiography in, for example, medical diagnosis results from

Figure 1.1: Schematic diagram of a conventional transmission radiography system. Drawn after Fig. 16.1 in Ref. [4].

the fact that in most cases different tissues have different attenuation coefficients and it is also often the case that when a tissue becomes diseased its attenuation properties change sufficiently to be identified in an x-ray image. In some cases, however, the change is less than the natural variation in attenuation for the healthy tissue and therefore not all disease can be detected using the information provided by transmitted x rays.

The development of radiography was in parallel with the development of the other classic x-ray discipline: x-ray diffraction. In 1912, experimental physicists Friedrich and Knipping under the direction of Max von Laue⁶ were the first to observe interference patterns produced by x rays transmitted through a crystal.⁷ Laue had incorrectly expected that fluorescence x rays emitted by the regularly spaced atoms in the crystal lattice should interfere, and it was later that year that W. H. Bragg and W. L. Bragg⁸ explained the diffraction patterns by proposing that the incident beam has a continuous spectrum and that the crystal

acts as a three-dimensional diffraction grating. The Braggs subsequently solved the structure of several crystals by examining their diffraction patterns and laid the foundations for x-ray crystallography. A variety of x-ray diffraction techniques have since been developed for the analysis of polycrystalline and amorphous materials.⁹ X-ray diffraction is a special case of coherent x-ray scattering and the diffraction pattern arises from the interference of x rays scattered from different sites in the crystal.

There was little dialogue between the two x-ray disciplines. On the one hand, x-ray diffraction studies are typically restricted to small, homogeneous samples and so the techniques cannot be applied to the extended, inhomogeneous objects, such as patients or suitcases, that are commonly imaged in radiography applications. The size restriction arises because for coherent scatter, the basis of diffraction, the angle through which photons scatter is related to their energy. As a result, diffraction techniques require nearly monochromatic radiation to give sharp patterns. The most practical monoenergetic source is the characteristic K radiation of x-ray tube targets, and diffraction procedures use metal targets with atomic number in the range 24–47, such as copper whose K_α line is 8 keV.⁹ Higher atomic number targets have higher energy characteristic radiation but are unsuitable because they produce very intense continuous bremsstrahlung radiation. Furthermore, lower energies are preferred because the interference peaks spread to larger angles, facilitating the acquisition and analysis of diffraction patterns. For these reasons, x rays of energy ~ 10 keV, which are unable to penetrate a few cm of even low density materials, are used in diffraction studies and specimens must be small. The homogeneity requirement arises because the diffraction pattern is a superposition of the scatter from all points of the object along the path of the primary beam.

On the other hand, coherent scatter was often ignored in radiography¹⁰ because higher energy x rays, typically continuous bremsstrahlung spectra from tungsten targets having photon energies spanning 20–150 keV, are utilized to penetrate thick sections of objects.

At these higher energies the total coherent scatter cross section is relatively small and the interference peaks get compressed to small angles, assumed to be so small that the elastic coherent scatter in radiography is indistinguishable from the unscattered primary beam.

Transmission CT imaging was widespread by the late 1970s and it was recognized by the early 1980s that the detection of scattered radiation in CT acquisitions caused cupping and streak artefacts.^{11,12} Subsequently, several groups performed calculations, which took into account intermolecular interference phenomena in the coherent scatter cross section, as well as measurements to study the distribution of scattered radiation in diagnostic imaging geometries.^{10,13,14} It was found that coherently scattered photons do diverge sufficiently from the primary beam to be separately detected and that the interference effects give low-angle structured peaks of intensity whose angular positions are material dependent. Furthermore, at small scatter angles ($<10^\circ$) the intensity of single coherent scatter is much more intense than single Compton scatter, despite the reverse being true for the total cross sections. Monte Carlo simulations¹⁵ confirmed these results and also showed that for a pencil beam, single coherent scatter dominates over multiple scatter at low angles, even for thick objects.

These facts suggested that it was possible to devise a system for measuring the diffraction properties of an extended object using a radiographic x-ray beam, that is, to join the techniques of x-ray diffraction and x-ray radiography. Harding¹⁶ and co-workers at Philips Research Labs in Hamburg were the first to do so in 1985, having developed a coherent scatter x-ray tomography system. Such a system utilizes the information-rich coherent scatter that is always present but has otherwise been ignored in conventional radiography.

While x-ray transmission is sensitive to material density and effective atomic number, coherent scatter is sensitive to the molecular structure. The integration of x-ray diffraction and radiography formed a powerful imaging technique based on a contrast mechanism that allows material discrimination not possible with transmission imaging. For instance,

disease may cause changes at the molecular level, impairing tissue function, but without a change in overall density. The work of this thesis is concerned with the development of coherent scatter projection imaging systems. The remainder of this introductory chapter is a literature review of coherent scatter imaging systems developed since 1985, followed by further motivation for the work completed for this thesis and how it advances the field.

1.1 Review of coherent scatter imaging systems

1.1.1 The first scatter imaging systems

From the outset, coherent scatter imaging (or x-ray diffraction imaging, as it is called by some authors) systems were tomographic or volumetric, measuring the diffraction profiles of an object on a voxel-by-voxel basis. Harding et al.^{16–18} originally developed a reconstructive coherent scatter CT system (Fig. 1.2). The beam from an x-ray tube operated at 120–140 kVp was collimated to a 1.0 mm diameter pencil to unambiguously distinguish primary from scatter intensity at the detection plane. A 15 cm thick plastics object was translated through the beam to acquire data for one projection. After each translation pass the object was rotated by 1.5°, and multiple projections were acquired. Discrete energy-integrating Bi₄Ge₃O₁₂ crystals (up to 61 in a 2D array) subtending an angular range of 1° simultaneously measured the transmitted and scattered x rays at different angles with respect to the midplane of the object. An ensemble of images with 1 mm pixels was reconstructed with filtered backprojection for a single slice of the object, one image for transmission and one for each scattering angle. Plastics appearing identical in the transmission image could be distinguished using the scatter measurements at different angles. The stationary anode x-ray tube was operated continuously at 5 mA and the data for the image set were acquired in 30 min.

The coherent scatter cross section is a function of x , the momentum transfer argument

Figure 1.2: Schematic diagram of a reconstructive coherent scatter CT system. Drawn after Fig. 2 in Ref. [16].

(defined below in Sec. 2.2.2), which links photon energy and scattering angle. It was recognized from early on that peaks in the measured scatter profiles are broadened by the spectral width of a polychromatic source and due to the thickness of the object, since photons originating from different points along the pencil beam need to scatter at different angles to reach the same point at the detector plane.^{15,17} Maximizing contrast between materials requires good momentum resolution, that is, being able to determine the energy and angle of scatter of detected photons.^{17,19}

The Philips group subsequently developed an energy-dispersive direct tomography (DT) scatter imaging system (Fig. 1.3) using a polychromatic x-ray pencil beam (100 kVp) and an energy-resolving high-purity germanium (HPGe) detector.^{20,21} Collimators between the object and detector permitted only 3.6° scatter to reach the detector. Knowing the angle and measuring the energy of each scattered photon, the scatter intensity could be computed as a function of momentum transfer. It is direct tomography because the detected scatter can

be directly mapped to a restricted region in the object (the intersection of the detector acceptance and pencil beam volumes) without reconstruction from multiple projections. The single detector system interrogated one voxel at a time and a 2D raster scan was needed to acquire an entire image slice (in any plane). The voxel was 5 mm in the depth dimension and at a fixed distance in front of the detector. Depth and angle resolution was at the expense of total scatter intensity and measurements took 100 s per voxel. A set of images for an object slice transverse to the beam was obtained by integrating scatter intensity over different momentum transfer ranges.

Figure 1.3: Schematic diagram of a single-voxel energy-dispersive DT coherent scatter imaging system. Drawn after Fig. 1 in Ref. [22].

1.1.2 Implementation variations

Following the pioneering work of Harding et al., several groups developed variations of the reconstructive CT and single-voxel DT systems in order to improve the accuracy and

resolution of measured scatter profiles or to acquire measurements in a shorter time. Many such systems were not designed to generate images but only to measure the scatter profile at one point for material characterization.^{22–40} The systems can be energy dispersive, using a polychromatic source and measuring the energy of scattered photons, or angle dispersive, using a monochromatic source and measuring the scattering angle. Monochromaticity is not a strict requirement for angle dispersive measurement and heavily or even lightly filtered spectra can be used at the expense of angular blur.

Royle and Speller^{23,24} developed a HPGe-based angle- and energy-dispersive single-voxel DT setup using slit collimation to give a larger voxel and higher scatter intensity. The angle-dispersive DT system of Evans et al.²⁵ used a filtered beam from a copper anode x-ray tube and a multi-wire proportional counter. Ndlovu et al.²⁶ used radionuclide sources and positioned a single HPGe detector to measure scatter at different angles. Newton et al.²⁷ used copper K_{α} x rays and x-ray film. Grant et al.⁴¹ developed an angle-dispersive reconstructive CT system using a sub-mm pencil beam of silver K_{α} x rays and a NaI detector. They were also the first develop a coherent scatter CT system using a monochromatic synchrotron beam, doing so at the Photon Factory synchrotron in Japan.⁴² Hall et al.⁴³ developed an energy-dispersive single-voxel DT system using a $50\text{ }\mu\text{m} \times 50\text{ }\mu\text{m}$ white beam at the ESRF synchrotron. Systems were developed using a number of other synchrotron facilities.^{34,44–48} Beevor et al.⁴⁹ used a tungsten anode x-ray tube, employing the balanced-filter technique (erbium/thulium)⁹ to achieve a nearly monochromatic beam, along with an x-ray image intensifier tube (XRII) and scatter collimation to admit scatter only from a fixed depth but at many scatter angles. Westmore et al.^{50,51} developed a reconstructive CT system using an XRII and a gadolinium-filtered tungsten anode pencil beam. For simplified data processing, Barroso et al.⁵² developed a reconstructive CT system with an energy-sensitive Si-Li detector to measure only photons having a single predetermined momentum transfer value.

Since the coherent scatter cross section is small and the scattered photons are spread over a large area, early scatter imaging systems acquired data too slowly or with too much radiation dose for practical application. The geometry of a system needs to be optimized for specific imaging tasks and materials.^{24,53} For energy-dispersive systems the optimal scattering angle was studied.^{22,24,54} Luggar et al.²⁸ determined that object thickness about two-thirds of the photon mean free path in the material maximizes scatter intensity. Geometries with relaxed angle and voxel resolution were studied.⁵⁵ Computational⁵⁶ and experimental⁵⁷ methods have been developed for application-specific optimization. The high cost and cooling requirements of HPGe detectors led to investigation of room temperature operated mercuric iodide⁵⁸ and CdZnTe^{59,60} detectors. Energy-dispersive and angle-dispersive systems have been compared.^{23,61} Sources of angular blur in angle-dispersive systems were investigated^{17,50} and a dominant effect when using a filtered x-ray tube beam is spectral blur. Reduced blurring was demonstrated when using the balanced-filter technique.⁶² Luggar et al.⁶³ modelled sources of profile blur in the energy domain.

1.1.3 Scatter image data processing and analysis

Complementary to system implementation, a variety of data processing and analysis techniques were also investigated to improve scatter profile accuracy as well as material identification and quantification. Reconstruction algorithms were developed for CT systems.^{64–66} Zhu et al. studied reconstruction with limited projections⁶⁷ as well as quasi-interior tomography.⁶⁸ Attenuation effects were corrected using photodensitometry techniques, scatter peak ratios,⁶⁹ multivariate calibration,⁷⁰ and normalization by transmitted intensity.⁷¹ Royle and Speller³⁰ acquired calibration data to determine the relative concentrations of two-component mixtures. For objects composed of a finite number of known materials, material-specific scatter images could be generated with reference to pure material basis functions by measuring scatter intensities for different ranges of momentum transfer

and solving linear equations⁷² or by non-negative least squares.^{51,73} Kleuker et al.⁴⁵ derived material-specific weighting factors prior to CT reconstruction and performed one reconstruction for each material. Material concentrations were also derived using multivariate techniques,^{70,74} singular value decomposition,³² and data clustering.⁴⁶ Madden et al.⁷⁵ used neural networks, wavelet transforms, and support vector machines to detect the presence of a target material. O’Flynn et al.⁷⁶ used principal component and discriminant analyses. Schlesinger and Bomsdorf⁷⁷ used a deconvolution method to improve spatial resolution of single-voxel DT. Van Uytven et al.⁷⁸ developed an algorithm to obtain the 3D electron density distribution from measurements of single Compton and coherent scatter from one projection.

1.1.4 Technological advances

Reconstructive CT

The need for fast acquisition of coherent scatter data also led to many technological advances. Schlomka et al. simulated⁷⁹ and built a prototype^{80–82} fan beam coherent scatter reconstructive CT system to enable acquisition of a slice projection in one shot without translation. The initial system had a multi-element energy-integrating line detector oriented perpendicular to the fan to give angle-dispersive scatter measurement and the detector was moved along an arc to emulate multi-line acquisition for each projection. Thin tungsten vanes were placed in front of the detector to restrict scatter to have been from a single primary path. Projection data were fed into an iterative algebraic reconstruction algorithm. The fan beam system was progressively improved upon by introducing a single-pixel energy-resolving CdTe detector,^{83,84} developing a filtered backprojection reconstruction algorithm using curved backprojection trajectories,^{85,86} and introducing a multiple scatter correction method.⁸⁷ A multi-slit primary collimator focused on the transmission detector

was used in conjunction with a long-line-focus x-ray tube to give a tenfold increase in fan intensity.⁸⁸ An 8×8 multi-line CdTe detector was implemented to measure a larger portion of the scattered photons.⁸⁸ More recently Zhu et al.⁸⁹ demonstrated cone beam CT with illumination coding.

Direct tomography

Advances were also made with direct tomography systems. The single-voxel energy-dispersive DT system was later classified as first generation DT.⁹⁰ Second generation DT utilizes a linear array of energy-sensitive detectors and fixed-angle scatter collimation to simultaneously measure scatter from the line of voxels along the pencil beam.⁹⁰⁻⁹² A 2D raster scan transverse to the beam yields scatter profiles for all voxels in the 3D object. Currently under development are third generation DT systems⁹⁰ which simultaneously acquire scatter profiles from a 2D array of voxels. Translation in one dimension (e.g. on a conveyor belt) is sufficient for tomographic imaging of a 3D object.

Harding⁹² defined several parameters to describe generic geometries for third generation DT setups. A number of specific geometries are of interest. The parallel beam geometry had been proposed in an angle-dispersive setup using a filtered beam from a line source x-ray tube and a flat-panel detector with orthogonal collimators⁹³ and in an energy-dispersive setup using a 2D array of energy-resolving detector elements with a multi-focus x-ray tube.⁹⁴ A direct fan beam geometry has the largest photon throughput per unit source power but requires a large 2D energy-resolving detector array and complicated scatter collimation with angled interlocking vanes. Tunna et al.⁹⁵ used novel manufacturing techniques to make a prototype collimator and Cui et al.⁹⁶ designed a fan beam system using a polycapillary array. Harding et al.⁹⁷ developed a test bed system to experimentally compare different third generation DT geometries and concluded that the inverse fan beam topology has the most promise considering photon throughput, available technology, and multiple

scatter effects. Radiation source⁹⁸ and detector⁹⁹ considerations were studied for inverse fan beam DT. A multiple inverse fan beam topology is being developed.^{94,100–102}

Coded apertures

Use of coded apertures as an alternative to the heavy collimation of DT systems was proposed early on.^{18,103} A group at Duke University^{104–106} was the first to implement coded aperture coherent scatter imaging in 2012 (Fig. 1.4). Coded masks between the object and detector attenuate less scatter than DT collimation and scatter from voxels at different positions project unique shadows of the aperture onto the detector. Volumetric data can be acquired without object rotation but the recorded scatter cannot be directly mapped to voxels and postprocessing of the multiplexed signal is needed. Different coding strategies have been investigated^{107,108} and improvements have been made to the postprocessing algorithm.¹⁰⁹ Multi-view acquisition with coded apertures has been studied.¹¹⁰ Fan beam systems for single-shot acquisition of a 2D image slice have been demonstrated.^{111,112} Full-volume single-shot acquisition with a cone beam has been simulated.¹¹³ The imaging performance of coding systems has been compared with DT systems.¹¹⁴ Machine learning techniques have been studied for material classification.^{115,116}

Figure 1.4: Schematic diagram of a coded aperture coherent scatter imaging system. Drawn after Fig. 1 in Ref. [114].

Focal construct geometry

Another recent development in coherent scatter imaging is the focal construct geometry of Dicken et al.^{117–124} An x-ray beam is collimated to a conical shell and cones of scatter originating from an annulus-shaped irradiated region of the object converge to regions of high intensity at the detector. Depth resolution can be achieved by use of a post-object circular diaphragm.¹²³ A 2D raster scan of the conical shell yields a volumetric image.¹²⁴

1.1.5 Applications

Clinical applications

The development of new imaging systems is driven by their potential for application, which for coherent scatter systems is nearly any situation that requires discrimination of similar density materials. Bone imaging was the first to be studied.^{21,125} The quantity of bone mineral and fat in just the trabecular bone can be determined from coherent scatter measurements and the results were verified by ashing and histomorphometry.²⁷ Further investigations with bone-mimicking phantoms,^{26,70} excised samples,^{23,24,26,30,126–128} and archaeological bone⁶⁹ demonstrated the utility of coherent scatter imaging for diagnosing bone demineralization diseases. Batchelar et al.^{129,130} generated material-specific tomographic images mapping the density distributions of the predominant bone constituents.

Breast imaging^{25,34,35,46,48,66,78,131–138} and urinary stone analysis^{31,139–141} are two other clinical applications that have garnered interest. Conventional screening mammography in the United States for women aged 50–69 is about 92% sensitive and 83% specific for initial screens,¹⁴² which leads to many unnecessary biopsies. Adipose, fibrous, glandular, and cancerous tissues can be classified using coherent scatter signatures. Treatment of urinary stones is influenced by their composition and some can be treated noninvasively with lithotripsy. Since coherent scatter is sensitive to chemical composition, it has the

potential for preoperative identification of stone type.

Security applications

Security was also one of the earliest areas of application to be identified²⁰ and continues to be of major interest today. In particular, much has been published on the screening of airline baggage for explosives and narcotics.^{22,29,32,33,49,53,60,76,94,103,143–147} Common plastic and inorganic explosive materials have density similar to harmless materials leading to many false alarms with conventional radiography systems, but can be easily identified by the distinctive peaks in their diffraction patterns. By 1991 the FAA was funding a program to develop an airport scanner.¹⁴⁸ The first airport prototype was made by Philips Research Labs⁹¹ and tested at the Hamburg airport.¹⁴⁹ L-3 Communications tested a prototype at the Orlando International Airport.⁷⁵ To speed up inspection, they employed a two-tier system, analyzing with coherent scatter only targeted volumes identified as suspicious by a conventional scan.

Other applications

Other applications of coherent scatter imaging that have been investigated include food-stuff contamination detection,^{28,72,150,151} gallstone disease management,²⁴ classification of colorectal liver cancer,³⁸ identification of pigments in historical paintings,¹⁵² imaging of atherosclerotic arteries,¹⁵³ and liquids identification.^{36,39,102,119,154,155} Tumour identification, liver and kidney deposits, and brain metabolism have been proposed as medical applications.²⁴ Proposed industrial and security applications include plastics sorting for recycling,^{151,156} screening for illicit materials in postal items,^{28,37,54,157} quality control of chemical substances in pharmaceutical and mineralogical industries,⁵⁸ and measuring heavy metal concentration in human tissues due to environmental or industrial exposure.⁴⁷

1.1.6 Modelling imaging performance

The first model to systematically analyze the contrast and SNR in scatter imaging was formulated by Leclair and Johns,^{158,159} following the approach of Motz and Danos¹⁶⁰ for modelling transmission imaging. The image quality metrics were computed by numerical integration for a variety of imaging tasks and it was shown that scatter imaging can outperform transmission imaging in a number of situations. The model was extended to account for the effects of a polychromatic source¹⁶¹ and was validated experimentally.¹⁶² The model was used to identify ranges of photon momentum transfer that maximize SNR.¹⁶³

1.1.7 Measurement of cross sections

Model calculations and computer simulations are valuable tools for the design and optimization of coherent scatter imaging systems but they rely on accurate coherent scatter differential cross section data. The cross section is proportional to the square of a form factor which scales the scattered intensity to account for intra- and inter-molecular interference effects between scatterings from different electrons (see Sec. 2.2.2). It can be computed from first principles for isolated atoms and simple molecules using Hartree Fock wave functions but the theoretical methods become intractable for amorphous assemblies of molecules and the form factor must be measured experimentally.¹⁶⁴ In the mid-1980s data were only available for water.¹⁶⁵ Seminal measurements for tissues and plastics were published by Kosanetzky et al.¹⁶⁶ Other groups looked at a restricted range of materials.^{30,35,131,167–171} In previous work by our group, angle dispersive cross section measurements were made with diffractometers¹⁶⁴ and with a computed radiography plate.¹⁷² Diffractometers are designed for precisely locating sharp peaks and were found to be unsuitable for measurements on amorphous materials. Among the more recent form factor data are those of King et al.¹⁷³ which were obtained with a more accurate energy dispersive method using a HPGe

detector.¹⁷⁴

1.1.8 Other imaging technologies

Although beyond the scope of this work, it would be remiss not to mention two other widely used imaging and analysis techniques that utilize x-ray forward scatter but are distinct from coherent scatter imaging: phase-contrast x-ray imaging (PCI) and small-angle x-ray scattering (SAXS). The three techniques are sensitive to molecular structure at different length scales. PCI^{2,175,176} is concerned with structure at scales larger than 1 μm which affect the refractive properties of materials. Some molecules such as collagen are highly ordered up to length scales of tens of nm and this structure is probed by SAXS.¹⁷⁷ Structure at this scale gives diffraction peaks at very small scattering angles ($<1^\circ$) and the technique requires microbeams from high intensity x-ray sources. Structure on the Angstrom level is the domain of x-ray diffraction (coherent scatter) studies. It should be noted that there is some overlap between SAXS and coherent scatter and some authors refer to coherent scatter imaging studies as SAXS or low-angle x-ray scattering.

1.2 Coherent scatter projection imaging

To summarize, nearly all x-ray coherent scatter systems have been voxel-based volumetric imaging or point measurement systems that can be grouped in three classes: reconstructive CT, direct tomography, and coded apertures. There has, however, been very little dedicated study of scatter *projection* imaging systems,^{18,153} even though they are routinely used for transmission x-ray imaging. Additionally, some single-voxel DT setups were effectively projection imaging since the samples used were thinner than the length of the voxel.^{72,150,151} Although projection imaging has no spatial resolution along the projection dimension it has some advantages over the volumetric imaging systems. Projection

imaging is technically simpler, not requiring object rotation/360° access around the object or complicated scatter collimation. Dispensing the need for multiple projections (vs reconstructive CT) and for discarding usable scatter intensity (vs DT and coded aperture systems) means image acquisition has the potential to be faster and that higher SNR can be achieved for the same object dose. For these reasons, an x-ray scatter projection imaging research program has been carried out in Paul Johns' group at Carleton University.

A simple scatter projection imaging system using a filtered x-ray tube beam and an energy-integrating computed radiography plate was demonstrated by Mohammad Nisar.¹⁷⁸ Seven pencil beams in a hexagonal arrangement simultaneously irradiated the object to reduce measurement time and to increase utilization of source output. The beams were far enough apart so that there was little overlap of scatter patterns at the detector. Exposures were tens of seconds long to record scatter patterns of sufficient intensity. The readout procedure of the image plate was very time consuming.

To circumvent the issues of low intensity and beam polychromaticity, the next stage of projection imaging development was carried out by Karl Landheer¹⁷⁹ at the Bend Magnet beamline of the Biomedical Imaging and Therapy (BMIT) facility at the Canadian Light Source (CLS) synchrotron.¹⁸⁰ A line of five 33.17 keV parallel pencil beams simultaneously irradiated the object, giving overlapping scatter patterns at the flat-panel x-ray detector. An iterative Maximum Likelihood-Expectation Maximization algorithm was used to extract the individual scatter profiles associated with each beam.¹⁸¹ The transmitted x rays were recorded simultaneous to the scatter using a set of discrete photodiodes. Translation stages scanned the object through the beams to acquire image data in a step-and-shoot manner. The fastest image acquisition time was 56 min for a 78.3 mm × 78.9 mm object several cm thick with 2900 pixels in the image. Images of a tissue phantom had fat-muscle contrasts of 0.10 ± 0.01 and 1.16 ± 0.03 for the primary and scatter images, respectively, demonstrating the contrast improvement achievable with a projection geometry for coher-

ent scatter imaging, but the imaging time was a limitation.

1.3 Hypothesis and objectives

The modelling work described in Sec. 1.1.6 predicted coherent scatter x-ray imaging to outperform conventional imaging in terms of contrast and SNR for a variety of imaging tasks. This has been demonstrated experimentally with the many imaging systems identified in Sec. 1.1 and, in particular, with the synchrotron-based projection imaging system described on the previous page. The proof-of-principle synchrotron system was slow to acquire data but had much potential for faster acquisition.

The hypothesis of this thesis is that a projection imaging system can be implemented to image a roughly $10\text{ cm} \times 10\text{ cm}$ area in less than ten minutes and provide soft-material contrast superior to conventional x-ray images. Imaging times shorter than ten minutes would place scatter projection imaging in the range of nuclear medicine scan times¹⁸² and therefore demonstrate technical feasibility for the clinical applications of coherent scatter imaging identified in Sec. 1.1.5.

The first goal is to build on Karl Landheer's prototype synchrotron-based system in order to maximize scatter projection image acquisition speed. Associated objectives are to measure the imaging performance of the system and to assess the effect of increased acquisition speed on image quality. The synchrotron provides an idealized environment for scatter imaging and bringing the system to maturity is a development stage of projection imaging. In addition to short acquisition time, a practical scatter imaging system needs to use a commonly available x-ray source, and so the second goal is to design and implement an x-ray tube-based system that can acquire a scatter image in less than ten minutes. Associated objectives are to measure the imaging performance of the x-ray tube-based system and to compare the performance to that of the synchrotron-based system.

1.4 Outline of thesis

In this work, I have developed two coherent scatter projection imaging systems and characterized them in terms of image quality. Chapter 2 presents the basic theory for the production of x rays as well as their interaction with matter, including details of the coherent scatter phenomenon, and the detection of x rays.

The first system uses a synchrotron source and is an improved version of Karl Landheer's proof-of-concept system. The major improvement was acquisition with continuous object motion in addition to using multiple pencil beams. The experimental setup, data acquisition procedures, data correction, image reconstruction, and methods for image quality assessment are presented in Chapter 3. Images and image quality quantification results are in Chapter 4.

A synchrotron source is very expensive and has limited access. In a move toward practical application of coherent scatter projection imaging, a second system using a standard laboratory x-ray source, a rotating anode x-ray tube, was developed. The challenges of using an x-ray tube source, system design optimization, technical details of the implemented x-ray tube-based system, and system assessment methods are given in Chapter 5. Images and other results are in Chapter 6. The conclusions of this thesis and avenues for future work are stated in Chapter 7.

1.5 Statement of contribution

This thesis is a summary of the author's research works completed while a doctoral student at Carleton University. The methods and results presented here have also been presented in peer-reviewed papers, conference proceedings, and conference presentations.

Paul C. Johns was the supervisor of the research projects and provided input for all aspects of the work in this thesis. The synchrotron-based system was the continuation of Karl

Landheer's M.Sc. research. George Belev was a beamline scientist at BMIT and provided suggestions for experimental setup and technical assistance for the synchrotron-based system. He wrote the LabVIEW code for that system and helped to revise peer-reviewed paper 1. Summer students Chris Dedek, Joey Carter, and Kelly Foran's investigations with the Perkin Elmer flat-panel detector guided design decisions for the x-ray tube-based system. Tong Xu was consulted about technical aspects of the x-ray tube-based system. MITACS intern Shuhang Chen performed most of the work for conference presentation 2.

I was responsible for writing all other code for simulation, control of the x-ray tube-based system, and data analysis, as well as for planning and executing experiments, for designing and building the x-ray tube-based system, for preparing and revising manuscripts and conference abstracts, and for preparing and giving conference oral presentations and posters.

1.5.1 Peer-reviewed papers

1. **Dydula, C.**, Belev, G. and Johns, P. C., Development and assessment of a multi-beam continuous-phantom-motion x-ray scatter projection imaging system. *Rev. Sci. Instrum.* **90**(3), 035104 (2019). Ref. [183].

The contents of this publication form the majority of Chapters 3 and 4.

2. **Dydula, C.** and Johns, P. C., Development and assessment of an x-ray tube-based multi-beam x-ray scatter projection imaging system. In preparation.

The contents of this publication form the majority of Section 5.4 and Chapter 6.

1.5.2 Conference proceedings

1. **Dydula, C.**, Belev, G. and Johns, P. C., Accelerated x-ray scatter projection imaging using multiple continuously moving pencil beams. *Proc. SPIE* **10132**, 101324J

(2017). Ref. [184]. Proceedings for poster presentation at SPIE Medical Imaging, Orlando, FL, 11 – 16 February 2017.

The contents of this publication form part of Chapter 3.

2. **Dydula, C.**, Xu, T. and Johns, P. C., Design and development of a rotating-anode x-ray tube coherent scatter projection imaging system. *Proc. SPIE* **11404**, 1140408 (2020). Ref. [185]. Proceedings for oral presentation at SPIE Defense + Commercial Sensing, Virtual, 27 April – 8 May 2020.

The contents of this publication form the majority of Sections 5.1–5.3.

1.5.3 Conference presentations

C. Dydula was the presenting author for each presentation listed below.

1. **Dydula, C.**, Belev, G. and Johns, P. C., Development of synchrotron-based x-ray scatter projection imaging. Poster presentation at CAP Congress, Ottawa, ON, 13 – 17 June 2016.
2. Chen, S., **Dydula, C.** and Johns, P. C., Two methods for the display of coherent x-ray scatter projection images. *Med. Phys.* **44**(8), 4377 (2017). Ref. [186]. Poster presentation at COMP Annual Scientific Meeting, Ottawa, ON, 12 – 15 July 2017.

Images from this work are presented in Sec. 4.4.
3. **Dydula, C.** and Johns, P. C., X-ray scatter projection imaging of an ex vivo canine prostate. *Med. Phys.* **44**(8), 4378 (2017). Ref. [187]. Poster presentation at COMP Annual Scientific Meeting, Ottawa, ON, 12 – 15 July 2017.
4. **Dydula, C.** and Johns, P. C., A scanning method for x-ray flat-panel calibration. *Radiother. Oncol.* **129**(Suppl. 1), S47 (2018). Ref. [188]. Poster presentation at

CARO-COMP-CAMRT Joint Scientific Meeting, Montreal, QC, 12 – 15 September 2018.

Section 3.3.1 is an expanded presentation of this work.

5. **Dydula, C.** and Johns, P. C., Design study for multi-beam x-ray scatter projection imaging with pre-detector vanes. *Med. Phys.* **46**(6), e300 (2019). Ref. [189]. Oral presentation at AAPM Annual Meeting, San Antonio, TX, 14 – 18 July 2019.

This work is briefly discussed in Sec. 7.2.3.

6. **Dydula, C.** and Johns, P. C., Benchtop coherent x-ray scatter projection imaging: From synchrotrons to x-ray tubes. *Med. Phys.* **47**(6), e383 (2020). Ref. [190]. Oral presentation at Joint AAPM-COMP Meeting, Virtual, 12 – 16 July 2020.

Chapter 2

Theory

The general idea of the imaging systems developed in this work is to irradiate an object with beams of x rays and to measure the spatial distribution of x-ray intensity that emerges behind the object, from which its composition and other information about it are inferred. X rays are a form of electromagnetic radiation whose origin is the acceleration of charged particles in a magnetic field or in the electric field of another charged particle, or from the transition of an electron between the shells of an atom.¹⁹¹ As for all forms of electromagnetic radiation, x rays have associated with them a wavelength, λ , and a frequency, ν , and travel in vacuum with a speed $c = 2.998 \times 10^8 \text{ ms}^{-1}$. The energy of an x-ray photon is related to its frequency as

$$E = h\nu, \tag{2.1}$$

where $h = 6.63 \times 10^{-34} \text{ Js}$ is Planck's constant. Radiations with wavelengths roughly in the range 10 nm to 10 pm form the x-ray part of the electromagnetic spectrum.⁴ This chapter presents the theory of how x rays are produced, how they interact with matter, and the methods for their detection, as relevant to this work.

2.1 Production of x rays

Sources of x rays include radionuclides, astrophysical sources, x-ray tubes, synchrotron light sources, and free electron lasers. The coherent scatter projection imaging systems developed in this work are based either on a synchrotron light source or an x-ray tube source. The principles of x-ray production and the characteristics of these two sources are described in this section.

2.1.1 Synchrotron light sources

Reference [2] was used as a general reference for this subsection. Synchrotron radiation is the radiation emitted by relativistic charged particles when they are accelerated radially in an applied magnetic field. A synchrotron light source consists of a storage ring, in which bunches of electrons circulate at a constant energy, and beamlines, down which beams of x rays emitted by the electrons travel to be used in experiments.

The electrons are emitted from an electron gun and are accelerated to their circulating energy in a linear accelerator and in a booster ring prior to being injected into the storage ring. The storage ring consists of curved sections and straight sections. The curved sections contain bending magnets which direct the electrons around the storage ring. The straight sections may contain focusing magnets, radiofrequency (RF) cavities, or insertion devices, which are periodic arrays of magnetic poles that force the electrons to travel in an undulating trajectory. X rays are emitted when the electrons have their direction changed by the bending magnets or insertion devices. The RF cavities in the storage ring serve to replace the energy lost by the electrons to radiation.

Along the beamline, a monochromator selects a particular average wavelength and bandwidth of x rays by means of optical dispersion or diffraction. The beam can be focused using x-ray mirrors and lenses. After irradiating and interacting with a sample, the

radiation is measured with a detection system.

Bending magnet radiation

For bending magnet radiation, the photon beam has a continuous spectrum and is emitted in a narrow cone about the line tangent to the electron's trajectory. The opening angle of the cone is $1/\gamma$, where $\gamma = \mathcal{E}_e/(m_e c^2)$ is the Lorentz factor, $\mathcal{E}_e$ is the total energy of the electron, and $m_e = 511 \text{ keV}/c^2$ is the mass of the electron. When viewed in the orbit plane of the electrons, the radiation has a horizontal linear polarization. Above or below the orbit plane, the polarization is elliptical.

The central or characteristic frequency of the radiation is

$$\nu_c = \frac{3}{4\pi} \frac{\gamma^2 e B}{m_e}, \quad (2.2)$$

where B is the magnetic field strength of the bending magnet and e is the magnitude of the electron charge. The average energy of the emitted radiation is proportional to the square of the energy of the electrons circulating in the storage ring and is directly proportional to the field strength of the bending magnet. Figure 2.1 is a plot of the bending magnet photon spectral flux density in photons per second per mrad^2 solid angle per 0.1% photon energy bandwidth for $\mathcal{E}_e = 3 \text{ GeV}$, $B = 1.5 \text{ T}$, and a 250 mA electron current in the storage ring.

Insertion devices

An insertion device (Fig 2.2) can be inserted in a straight section of the storage ring to produce much more intense x-ray beams. It is an array of magnetic poles arranged in a periodic structure with spatial period λ_u such that the magnetic field direction alternates up and down, causing the electron to have an oscillating trajectory in the horizontal plane as it passes through the device. The device has N_{per} periods and $2N_{per}$ pole pairs. The electron

Figure 2.1: Photon spectral flux density for a 1.5 T bending magnet with a 250 mA current of $\mathcal{E}_e = 3$ GeV electrons circulating in the storage ring. The spectrum was computed using Eq. (27) in Ref. [192]. The characteristic photon energy is $h\nu_c = 9.0$ keV.

Figure 2.2: Schematic diagram of a synchrotron light source insertion device. The solid curve in the gap between the upper and lower sets of magnets is the trajectory of an electron bunch.

experiences a maximum magnetic field, B_0 , at the central position between a pole pair.

The amplitude of the oscillations in the insertion device is quantified by the electron's maximum angle of deflection from its straight trajectory, which is equal to K/γ , where¹⁹²

$$K = \frac{eB_0\lambda_u}{2\pi m_e c} \quad (2.3)$$

is the dimensionless undulator parameter. For K less than or of the order of one, the amplitude of oscillations is small and the insertion device is referred to as an undulator. For undulators, the radiations caused by deflections of the same electron by the different pole pairs add coherently but the radiation from different electrons is incoherent. The emitted spectrum is approximately monochromatic (with harmonics) and the full-width at half-max of the radiated intensity distribution about a harmonic frequency is proportional to $1/(nN_{per})$, where n is the harmonic number. The fundamental frequency radiated, when viewed along the undulator axis, is²

$$\nu_1 = \frac{2c\gamma^2}{\lambda_u} \left(\frac{2}{2 + K^2} \right), \quad (2.4)$$

and can be tuned by varying B_0 . The opening angle of the cone of undulator radiation is approximately equal to $1/(\gamma\sqrt{nN_{per}})$. The intensity of undulator radiation is 5 to 8 orders of magnitude greater than that of bending magnet radiation.²

For $K \gg 1$, the amplitude of oscillations of the electron trajectory is relatively large and the insertion device is referred to as a wiggler. For a wiggler, the radiation from the different pole pairs is incoherent and the device is effectively $2N_{per}$ bending magnets. The emitted spectrum is semi-continuous and the peak photon spectral flux density is $(2N_{per})^2$ times that of a single bending magnet with field strength B_0 .¹⁹² Since it radiates in a wide spectrum, a wiggler can output several orders of magnitude more photons per second than an undulator. For the same electron beam and number of pole pairs, wiggler output at the

fundamental frequency of an undulator can be similar or greater than the undulator output at that frequency. The opening angle of wiggler radiation is K/γ in the horizontal direction and $1/\gamma$ in the vertical direction.

Monochromators

The most common method to monochromatize the polychromatic x-ray beam from a bending magnet or insertion device is to use a crystal to diffract a specific wavelength, determined by Bragg's law, out of the incident beam. The diffracted beam has some spectral width and may have an increased angular divergence, depending on the geometry of the crystal. When the beam is diffracted by reflection it is referred to as a Bragg crystal. For a Laue crystal, diffraction occurs by transmission. Since the Bragg reflection changes the direction of the incident beam, a second crystal is often used to redirect the monoenergetic beam in the direction of the incident beam. Bending the crystals increases the flux and can focus or expand the size of the diffracted beam. Monochromator crystals used in synchrotron light source beamlines are typically silicon, diamond, or germanium.

2.1.2 X-ray tubes

Reference [4] was used as a general reference for this subsection. An x-ray tube is an evacuated glass or ceramic envelope containing a coiled tungsten filament (the cathode) and a metal target (the anode). A schematic is shown in Fig. 2.3. Heating the cathode filament by passing a current through it causes the thermionic emission of electrons from its surface. Applying a high positive voltage to the anode with respect to the cathode (between 20 kV and 150 kV for medical imaging) attracts these electrons to the anode. The inside of the glass envelope is a high vacuum so that the electrons do not collide with gas molecules and consequently they strike the anode with high speed. A focusing cup directs the electrons to a small area of the anode called the focal spot. The part of the anode that is bombarded

Figure 2.3: Schematic diagram of an x-ray tube. Drawn after Fig. 2.8 in Ref. [4].

with electrons is made of a tungsten-rhenium alloy or molybdenum for x-ray tubes used in diagnostic radiology.

The high voltage is supplied to the tube by the x-ray generator. The generator takes one or more phases of line voltage as input and uses a high tension transformer to step it up to the tube kilovoltage. It also has an isolation step-down transformer to step it down to about 10 V to energize the cathode filament. Ideally the voltage across the anode and cathode is constant for the duration of an exposure and, in order of decreasing voltage ripple, generators may be single-phase full-wave rectified (two pulses per cycle), three-phase with six rectifiers (six pulses per cycle), three-phase with twelve rectifiers (twelve pulses per cycle), or they may be high frequency generators (nearly constant voltage and compact in size). High frequency generators are the standard today.

For single- and three-phase generators, exposure times are an integer multiple of the

duration of a single pulse. In North America, where the line voltage is 60 Hz, the shortest exposure is $1/(60p)$ s, where p is the number of pulses per single-phase cycle. An x-ray exposure is characterized by the peak kilovoltage, kVp, which is the maximum of the pulsatile voltage applied to the tube, and by the mAs, which is the product of the average current, in mA, of electrons flowing from the cathode to the anode and the exposure time, in seconds. X-ray output is proportional to the mAs. The generator can provide tube currents from a few mA up to about 1 A.

When the high speed electrons strike the anode, x rays are emitted in all directions, at least half of which are absorbed in the target itself. The glass envelope is inserted in an oil-filled housing lined with lead, absorbing most of the remaining x rays, except for an opening at the tube port through which the useful cone of radiation is emitted. For imaging applications of x-ray tubes, the x rays should ideally originate from a single point on the anode to maximize image sharpness. In practice the focal spot has a finite area of about $1.0 \times 1.0 \text{ mm}^2$. About 10% of the radiation flux from the x-ray tube is off-focal radiation, that is, x rays emitted by electrons striking parts of the anode other than the focal spot. Off-focal radiation can be reduced by placing collimators as close as possible to the focal spot.

High energy electrons interacting with the anode can produce x rays in one of two ways. If the electron strikes an inner shell electron of an atom in the anode and its kinetic energy is greater than the binding energy, the bound electron will be ejected from the atom and an outer shell electron will transition to fill the vacancy in the inner shell. During the transition, a quantum of radiation is emitted from the atom, whose energy is equal to the difference in binding energy of the two shells involved. This is called characteristic radiation since the discrete energies of the photons emitted in this way depend on the atomic number of the target atom. Characteristic radiation is emitted isotropically and is unpolarized. The intensity of characteristic x rays from an x-ray tube is about ten orders of magnitude less

than that of an undulator synchrotron source.²

Alternatively, if an electron passes close to an atomic nucleus in the anode it will be deflected by its electric field and emit a bremsstrahlung x ray. The emitted photon can have energy ranging from zero up to the kinetic energy of the incident electron. For a beam of monoenergetic electrons incident on a thick target, such as an x-ray tube anode, the relative intensity of x rays produced per photon energy interval decreases linearly (Kramers' spectrum) from a maximum at 0 keV to zero at the energy of the incident electrons, which is determined by the peak kilovoltage across the tube. As the x-ray beam emerges from the target, through the exit window of the tube, and any added filters, the lower energy part of the spectrum is preferentially absorbed (discussed below in Secs. 2.2.4 and 2.2.5). Filters are added because the low energy x rays increase the dose but are unable to penetrate the object and be measured. The total x-ray intensity is proportional to the first power of the atomic number of the target and is proportional to a power higher than 2.0 of the kVp, when taking filtration into account. Thick target bremsstrahlung is unpolarized and, for diagnostic tube voltages, would be emitted isotropically if there was no absorption. An x-ray tube spectrum generated from a computational model is shown in Fig. 2.4. The continuous part is due to bremsstrahlung and the superimposed peaks are characteristic radiation.

For tube voltages used in medical imaging, a relatively small number of the electron interactions in the anode give rise to characteristic and bremsstrahlung radiation. More than 99% of the energy of the electrons is lost in interactions that lead to heating of the anode. If the thermal energy stored in the focal track area exceeds some critical value, the surface of the anode will melt and the x-ray tube will be destroyed. Therefore, there is a maximum power input with which the focal spot can be loaded and a limit on the number of successive exposures if the average rate of heating is faster than the rate of cooling. The focal spot loses thermal energy by conduction to the rest of the anode body and by infrared radiation to the walls of the tube, and the rate of cooling increases with

Figure 2.4: Tungsten anode x-ray tube spectrum at 75 cm from the source for constant tube voltage 110 kV and 2.5 mm of added Al filtration. Spectrum data from Ref. [193].

temperature. Tube manufacturers state maximum heat loads and provide cooling curves for the anode body and the tube housing. The tube housing has more heat capacity than the anode but dissipates heat more slowly by convection to the atmosphere.

Although a smaller focal spot improves image sharpness, the area of the anode bombarded by the electrons cannot be too small due to power input restrictions. There are two ways to increase the area over which electrons strike the anode but still maintain a small apparent focal spot. The target area of the anode can be made to slope away from the electron beam at a 6° – 16° angle between the target face and the plane perpendicular to the electron beam (Fig. 2.5). When viewed along the central axis of the cone of radiation, the length of the anode target area struck by the electrons appears to be reduced by the sine of the anode angle (line focus principle). The focal spot will appear smaller or larger when viewed from the anode or cathode side of the central axis, respectively. The intensity of the x-ray beam decreases toward the anode side due to absorption in the target (known as the heel effect). The anode can also be constructed as a disc with a diameter of about 10 cm and attached

Figure 2.5: A line focus rotating anode with anode angle ϑ . By design, when viewed along the central axis of the beam, the length of the focal spot appears to be reduced to be equal to its width, cd . The anode rotates with angular velocity ω . Drawn after Fig. 2.7 in Ref. [4].

to the rotor of an induction motor which spins at 3000 rpm (low speed) or 9000 rpm (high speed). This construction causes the electrons to strike a target at the same position relative to the tube housing but to be spread over a large area track around the anode disc. For a rotating anode tube, the stator coils of the induction motor must first be energized to bring the rotor up to its operating speed and then the high voltage can be applied to make an exposure.

2.2 Interaction of x rays with matter

An x-ray photon passing through an object can either be completely absorbed, exit the object without undergoing any interaction, or undergo one or more scattering events prior to being absorbed or exiting the object. A scattering event is one in which the photon changes direction upon interacting with a particle in the object and during which it may or may not

Figure 2.6: Geometry for the scattering of a photon by a free electron in its rest frame. The incident and scattered photons have energy $h\nu$ and $h\nu'$, respectively. The scattered electron has kinetic energy T .

have a change in energy. In the diagnostic energy range (roughly 20 keV to 150 keV), one of three processes occur for nearly all interactions of x rays with matter:⁴ coherent scatter, incoherent scatter, or photoelectric absorption. In this section these processes are discussed in some detail.

2.2.1 Scattering of a photon by a free electron

Consider first a photon with energy $h\nu$ incident on a free electron in its rest frame (Fig. 2.6). The incident photon travels in the z -direction and has some polarization, with the direction of the electric vector, $\hat{\epsilon}$, perpendicular to the direction of travel. After scattering, the photon has energy $h\nu'$ and travels in a direction making an angle θ , the scattering angle, with respect to its initial direction of travel. The electron acquires a kinetic energy T and travels in a direction making an angle ϕ with respect to the incident photon's direction of travel. The scattered photon has polarization vector $\hat{\epsilon}'$ and ψ is the azimuthal angle of scattering, that is, the angle between $\hat{\epsilon}$ and the projection of the scattered photon's direction of travel onto the plane perpendicular to the z -direction. The direction of $\hat{\epsilon}'$ is the direction of $\hat{\epsilon}$ projected onto the plane perpendicular to the direction of travel of the scattered photon.

Following from the conservation of relativistic kinetic energy and momentum, the energies after scattering are⁴

$$T = h\nu \frac{\alpha(1 - \cos \theta)}{1 + \alpha(1 - \cos \theta)}, \quad (2.5)$$

and

$$h\nu' = h\nu \frac{1}{1 + \alpha(1 - \cos \theta)}, \quad (2.6)$$

where $\alpha = h\nu/m_e c^2$. It can be seen in Eqs. (2.5) and (2.6) that if the energy of the incident photon is small compared to $m_e c^2$, or if θ is small, then the electron acquires a negligible amount of energy and the scattering is elastic ($h\nu \approx h\nu'$). In this limit with no electron recoil, a classical electromagnetic model may be used to derive the differential scattering cross section, which is defined to be the power radiated by the electron into a solid angle element $d\Omega$ divided by the magnitude of the energy flux of the incident x-ray beam, per unit solid angle. The result is²

$$\frac{d_e \sigma}{d\Omega} = r_0^2 |\hat{\mathbf{e}} \cdot \hat{\mathbf{e}}'|^2 = r_0^2 (1 - \cos^2 \psi \sin^2 \theta), \quad (2.7)$$

where $r_0 = 2.81794 \times 10^{-15}$ m is the classical electron radius. This is the Thomson differential scattering cross section for a polarized beam of x rays. For an unpolarized beam, averaging over the azimuthal angle, ψ , gives

$$\frac{d_e \sigma_0}{d\Omega} = r_0^2 \frac{1 + \cos^2 \theta}{2}. \quad (2.8)$$

As the energy of the photon increases and α is no longer negligible, a relativistic quantum model is needed to account for the dynamics of the scattering when there is electron

recoil. The result is the Klein-Nishina differential cross section,⁴

$$\frac{d_e \sigma_{KN}}{d\Omega} = \frac{d_e \sigma_0}{d\Omega} F_{KN}(h\nu, \theta), \quad (2.9)$$

where

$$F_{KN}(h\nu, \theta) = \left(\frac{1}{1 + \alpha(1 - \cos \theta)} \right)^2 \left(1 + \frac{\alpha^2(1 - \cos \theta)^2}{(1 + \alpha(1 - \cos \theta))(1 + \cos^2 \theta)} \right). \quad (2.10)$$

This factor is always less than 1.0, reducing the probability of scatter. As the energy of the incident photon decreases, $F_{KN}(h\nu, \theta)$ approaches 1.0 and the Klein-Nishina differential cross section reduces to $\frac{d_e \sigma_0}{d\Omega}$, the Thomson differential cross section. The process of photon scattering by a free electron at incident energies and scattering angles for which the electron recoil is important is known as Compton scatter.

2.2.2 Coherent scatter

Under usual laboratory conditions, the electrons in an object are not free but are bound to atoms or molecules. If the wavelength $\lambda = c/\nu$ of an x-ray photon is on the order of the size of an atom or molecule or of the average spacing between molecules in the object, then the physical event of the propagation of a photon from the source to the detector may occur by different alternatives, namely by being scattered by the different electrons in the atom, molecule, or collection of neighbouring molecules.¹⁹⁴ Quantum mechanically, there is a probability amplitude for each of these alternatives, and as long as a specific trajectory is not identified, the total amplitude for the scattering is the sum of the probability amplitudes of the alternatives. Since it is the same photon taking each trajectory, the scattered amplitudes are coherent. In this situation the recoil is taken up by the entire atom or molecule and since the masses of these systems are much greater than that of a single electron, the scattering

is effectively elastic and a term similar to $F_{KN}(h\nu, \theta)$ (Eq. (2.10)) can be omitted in the coherent scatter cross section. The coherent, elastic scattering of a photon is also called Rayleigh scattering.

Since the scattering electrons are at different positions, the probability amplitude for each trajectory has a different phase, depending on the path length via a particular electron to the detector. As a result, interference will be observed, which depends on the incident photon energy, the scattering angle, and the arrangement of the electrons. Since the probability of scatter from any electron is proportional to $\frac{d_e \sigma_0}{d\Omega}$, the probability amplitude for a trajectory is proportional to $\sqrt{\frac{d_e \sigma_0}{d\Omega}}$ multiplied by a phase factor $e^{i\varphi}$, where φ is the phase of the trajectory.

Define the scattering amplitude for the j th trajectory having phase φ_j to be

$$\Psi_j \equiv \sqrt{\frac{d_e \sigma_0}{d\Omega}} e^{i\varphi_j}. \quad (2.11)$$

To determine the total scattering amplitude, consider first the coherent scatter of a photon by two perfectly localized electrons, the first located at the origin and the second located at point P , having a position vector $\vec{r}$ relative to the origin (Fig. 2.7). The incident photon has wavelength λ and momentum $\vec{p}$, with magnitude $|\vec{p}| = h/\lambda$ and direction given by the unit vector $\hat{p}$. Since the scattering is elastic, the scattered photon has momentum $\vec{p}'$ with the same magnitude as the incident photon but with a direction $\hat{p}'$ making an angle θ with $\hat{p}$. As can be seen in Fig. 2.7, the path length difference between the two trajectories is

$$\Delta d = \vec{r} \cdot \hat{p} - \vec{r} \cdot \hat{p}' = -\vec{r} \cdot (\hat{p}' - \hat{p}), \quad (2.12)$$

and so the phase difference is

$$\Delta\varphi = \frac{2\pi}{\lambda} \Delta d = -\frac{2\pi}{\lambda} \vec{r} \cdot (\hat{p}' - \hat{p}). \quad (2.13)$$

Figure 2.7: Coherent scatter through an angle θ of an x-ray photon with wavelength λ by two electrons. One electron is located at the origin, O , and the second electron is located at point P .

Defining the momentum transfer argument as

$$\vec{x} \equiv \frac{\vec{\Delta p}}{2h} = \frac{\vec{p}' - \vec{p}}{2h}, \quad (2.14)$$

whose magnitude for elastic scatter is

$$|\vec{x}| = \frac{1}{\lambda} \sin(\theta/2) = \frac{E}{hc} \sin(\theta/2), \quad (2.15)$$

the phase difference may be written as

$$\Delta\phi = -4\pi\vec{r} \cdot \vec{x}. \quad (2.16)$$

If the phase of the scattering trajectory through the origin is taken to be zero, then Eq. (2.16) gives the phase of the probability amplitude for scattering by an electron at position $\vec{r}$. The total scattering amplitude can be computed by summing over the individual amplitudes, Ψ_j , for all trajectories involved but since individual electrons cannot be perfectly localized in an

atom, the electron density distribution (number of electrons per unit volume), $\rho_e(\vec{r})$, must be used.¹⁷⁷ This is the square modulus of the multielectron wave function of the electrons participating in the coherent scatter. The number of electrons in a volume element dV is equal to $\rho_e(\vec{r})dV$. The total scattering amplitude is then a volume integral,

$$S = \sqrt{\frac{d_e \sigma_0}{d\Omega}} \int_V \rho_e(\vec{r}) e^{-i4\pi\vec{r}\cdot\vec{x}} dV, \quad (2.17)$$

and

$$F_{coh}(\vec{x}, \rho_e(\vec{r})) \equiv \int_V \rho_e(\vec{r}) e^{-i4\pi\vec{r}\cdot\vec{x}} dV \quad (2.18)$$

is defined to be the coherent scatter form factor. By substituting $\vec{s} = 2\vec{x}$ in Eq. (2.18), it can be seen that F_{coh} is the Fourier transform of the electron density distribution. From the Fourier transform relationship it follows that the behaviour of F_{coh} at very small $|\vec{x}|$ is determined by long-range structure while large $|\vec{x}|$ behaviour ($|\vec{x}| > 2.5 \text{ nm}^{-1}$) is dominated by the core electrons of atoms which are unaware of how the atom bonds with its neighbours. The coherent scatter differential cross section is the square modulus of the total scattering amplitude,

$$\frac{d\sigma_{coh}}{d\Omega} = SS^* = \frac{d_e \sigma_0}{d\Omega} |F_{coh}(\vec{x}, \rho_e(\vec{r}))|^2. \quad (2.19)$$

For a crystalline object, the magnitude and direction of $\vec{x}$ need to be considered and the constructive interference of the coherent scatter produces an arrangement of dots of diffraction maxima (that is, the scatter depends on θ and ψ). If the object is amorphous, however, then the random orientations of molecules or local collections of molecules throughout the object wash away the coherent scatter dependence on ψ , giving rings of constructive interference maxima (Laue rings), similar to that for a powdered crystal but not as sharp. The interference observed will in general depend on the state of matter and density of the material. For amorphous objects, the coherent scatter form factor is a function of the magnitude

of $\vec{x}$, given by Eq. (2.15).

As θ approaches 0° , the phases for all of the trajectories become equal and there is perfect constructive interference. The coherent scatter form factor is then equal to its maximum value Z_{coh} , the number of electrons in the scattering volume or in the fundamental unit of the substance, and the coherent scatter differential cross section is enhanced by a factor of Z_{coh}^2 relative to the Thomson differential cross section. For scattering by a free neutral atom, $Z_{coh} = Z$, the atomic number. Accordingly, the differential cross section in Eq. (2.19) is a cross section per fundamental unit of substance (e.g. per atom or per molecule). It may be converted to a differential cross section per electron as,

$$\frac{d_e \sigma_{coh}}{d\Omega} = \frac{d_e \sigma_0}{d\Omega} \frac{|F_{coh}(\vec{x}, \rho_e(\vec{r}))|^2}{Z_{coh}}. \quad (2.20)$$

Sometimes in the literature^{164,173} the reported coherent scatter form factor is a form factor per electron, $_e F_{coh}(\vec{x}, \rho_e(\vec{r}))$, defined such that

$$\frac{d_e \sigma_{coh}}{d\Omega} = \frac{d_e \sigma_0}{d\Omega} |_e F_{coh}(\vec{x}, \rho_e(\vec{r}))|^2. \quad (2.21)$$

The coherent scatter form factor per electron is equal to the absolute value of the form factor in Eq. (2.18) divided by $\sqrt{Z_{coh}}$ and so it has a maximum value of $\sqrt{Z_{coh}}$.

For large $\vec{x}$ (large photon energy and/or scattering angle), the phase factors $e^{-i4\pi\vec{r}\cdot\vec{x}}$ vary rapidly with $\vec{r}$ and they average out to zero. The interference gets washed out and the coherent scatter becomes negligible.

2.2.3 Incoherent scatter

As the energy of the incident x rays increases and the wavelength decreases, in addition to the disappearance of coherent scatter, the probability increases for the x-ray photon to

interact with an individual electron. The binding energy of the electron becomes less significant compared to the energy of the incident photon and the recoil of the scattering is taken up by the individual electron. In this situation the Klein-Nishina differential cross section needs to be used. A collection of x-ray photons scattering off of individual electrons are incoherent and there is no interference. Since the electrons have a distribution of momenta relative to the rest frame of the atom, there is a Doppler broadening of the scattered photon energy. The Compton scatter of a photon by a bound electron is called incoherent scatter.

For a bound electron, there is some probability, given by the Klein-Nishina differential cross section, that the photon will scatter at an angle θ and the electron will acquire a kinetic energy T , as if it were free. There is then some probability less than 1.0 that this electron will actually leave the atom. Thus, the effect of the binding is to reduce the Klein-Nishina differential cross section. The incoherent scatter differential cross section is obtained from $\frac{d_e\sigma_{KN}}{d\Omega}$ by multiplying by a factor $S_{inc}(x, Z)$, called the incoherent scattering function:⁴

$$\frac{d\sigma_{inc}}{d\Omega} = \frac{d_e\sigma_0}{d\Omega} F_{KN}(h\nu, \theta) S_{inc}(x, Z). \quad (2.22)$$

Like $\frac{d\sigma_{coh}}{d\Omega}$ in Eq. (2.19), this is a differential cross section per atom. For large photon energies, for which the binding energy is negligible, or for large θ , $S_{inc}(x, Z)$ approaches its maximum value of Z . For small values of x , $S_{inc}(x, Z) \approx 0$. The incoherent scattering function per electron can be calculated as $S_{inc}(x, Z)/Z$ and has a maximum value of 1.0. Dividing Eq. (2.22) by Z gives $\frac{d_e\sigma_{inc}}{d\Omega}$, the incoherent scatter differential cross section per electron.

Plotted in Fig. 2.8 are $\frac{d_e\sigma_{coh}}{d\Omega}$, $\frac{d_e\sigma_{inc}}{d\Omega}$, and $\frac{d_e\sigma_{KN}}{d\Omega}$ as a function of θ for 60 keV x rays in water. The coherent scatter differential cross section is relatively large at low θ (small x) but decreases to zero for θ larger than about 15° . The shape of the curve is highly material-dependent. The incoherent scatter differential cross section is suppressed at low θ , at which

Figure 2.8: Coherent, incoherent, and Klein-Nishina differential scattering cross sections per electron per steradian for 60 keV x rays in water. Coherent scatter form factor and incoherent scattering function data from Ref. [173].

less energy is transferred to the electron, but approaches the Klein-Nishina differential cross section at higher θ . For higher energies, the coherent scatter curve retains nearly the same shape but is compressed to smaller angles until eventually all coherent scatter is constrained to very small angles about 0° . The incoherent scatter curve approaches the Klein-Nishina differential cross section more quickly as the binding energy becomes less significant at higher x-ray energies.

2.2.4 Photoelectric effect

The absorption of x rays in matter is due to the photoelectric effect. This interaction takes place with bound electrons. The incident photon with energy $h\nu$ is completely absorbed and the electron is ejected from the atom with kinetic energy $T = h\nu - E_b$, where E_b is the binding energy of the electron. Electrons having $E_b < h\nu$ do not participate in the

photoelectric interaction. The probability of interaction with a particular electron is greatest when the photon has just enough energy to eject the electron.

Exact theoretical determinations of the photoelectric cross section per electron, ${}_e\tau$, are difficult to obtain and the quantitative aspects are mostly empirical.¹⁹¹ The general dependence of ${}_e\tau$ on photon energy, $h\nu$, and on atomic number, Z , will be stated here. For $h\nu = 100$ keV, the cross section per electron varies approximately as Z^3 for high- Z materials and as $Z^{3.8}$ for low- Z materials. At 3 MeV the dependence goes approximately as $Z^{3.6}$ for high- Z materials. The dependence of ${}_e\tau$ on the photon energy has discontinuous jumps due to the sudden participation of inner shell electrons when $h\nu$ is greater than their binding energy. These jumps or absorption edges are insignificant for low- Z materials since the binding energies of the inner shell electrons are very small. Below about 100 keV, ${}_e\tau$ varies approximately as $(h\nu)^{-3}$ between the jumps. Above 1 MeV the dependence is closer to $(h\nu)^{-1}$. For low energies, the photoelectric effect is the predominant mode of x-ray interaction with matter but it becomes insignificant above 1 MeV.

2.2.5 Exponential attenuation of x rays

Integrating the differential scatter cross section per electron per steradian of a photon interaction with matter, $\frac{d{}_e\sigma_k}{d\Omega}$, over the entire solid angle gives the total cross section per electron, ${}_e\sigma_k$, for that interaction, where the index k denotes a particular interaction. The photoelectric cross section, ${}_e\tau$, is included in the set of total cross sections. Multiplying each ${}_e\sigma_k$ by the electron density (number per unit volume) of the material, ρ_e , gives the partial linear attenuation coefficient, μ_k , for that interaction type. The total linear attenuation coefficient is $\mu = \sum_k \mu_k$, which is the probability of interaction per unit length of material traversed by a photon. The linear attenuation coefficient for a material depends on the density of the material. The probability that a single photon has undergone at least one interaction after

Figure 2.9: Partial linear attenuation coefficients for the photoelectric, coherent scatter, and incoherent scatter interactions of 1–200 keV x rays in water with density 1.0 g cm^{-3} . Data from Ref. [195].

traversing a thickness t of a material is,

$$P = 1 - e^{-\mu t} . \quad (2.23)$$

Equivalently, for N_0 photons incident on an object, the number of photons having undergone no interaction after traversing a thickness t is,

$$N = N_0 e^{-\mu t} . \quad (2.24)$$

Plotted in Fig. 2.9 are the partial linear attenuation coefficients for x rays in water with density 1.0 g cm^{-3} as a function of photon energy. The photoelectric effect dominates at lower energies and incoherent scatter dominates at higher energies. The energy at which incoherent scatter begins to dominate is larger for higher- Z materials. For sufficiently low

energies, coherent scatter is more probable than incoherent scatter, however, for low- Z materials such as water, incoherent scatter takes over as the dominant mode of scatter at the lower end of the diagnostic energy range. For high- Z materials, coherent scatter becomes more probable over the most of diagnostic energy range, due to the Z^2 dependence of the coherent scatter cross section per atom.

2.3 Detection of x rays

After interacting with the object, the transmitted x rays have a spatial variation in intensity that depends on the distribution of materials within the object. The x rays downstream of the object then undergo interactions in a sensing device, constructed so as to allow for the measurement of the intensity distribution and for the display or storage of the x-ray image. The x-ray interactions lead to ionization of atoms and excitation of electrons to higher energy levels. X-ray detectors measure the electric charge liberated from ionization, measure photons in or near the visible range emitted by phosphors after excited electrons return to lower energy levels, or undergo chemical changes following ionization. Types of x-ray detectors include fluorescent screens, gas-filled ionization chambers, radiographic film, image intensifier tubes, scintillator crystals coupled to photomultiplier tubes, computed radiography plates, and digital flat-panel detectors. The latter were used in this work and their principle of operation will be described.

2.3.1 Flat-panel detectors

Reference [196] was used as a general reference for this subsection. A flat-panel x-ray imager is a two-dimensional digital detector that samples and discretizes the incident field of x rays in both the spatial and intensity dimensions. It consists of an x-ray detection medium, an active-matrix array, and peripheral electronics, and integrates the energy of the incident

Figure 2.10: Cross-sectional diagram of an indirect flat-panel detector. Drawn after Fig. 4.1 in Ref. [196].

x-ray beam over a finite time period rather than counting individual photons. Flat-panel imagers are classified as either indirect detectors, if the x-ray detection medium is a phosphor layer producing photons in the visible range upon x-ray interaction, or direct detectors, if the detection medium is a photoconductor producing electrical charges. Indirect flat-panel detectors (Fig. 2.10) were used in this work.

The x-ray detection medium serves to absorb x rays and produce a response at the surface that is localized to where the absorption occurred and that is proportional to the energy absorbed. X-ray absorption promotes electrons from the valence band to the conduction band of the medium. Phosphors have a wide band gap and the electrons return to the valence band via local states in the forbidden gap created by activator impurities, emitting light in the process. The phosphor layer of an indirect detector is either a fluorescent screen consisting of a powdered phosphor with a high refractive index, such as $\text{Gd}_2\text{O}_2\text{S:Tb}$, mixed with a transparent binder, or a structured phosphor scintillator, such as CsI:Tl . The physical properties of phosphor materials include the number of light photons emitted per unit energy of absorbed x rays, the wavelength of the emitted photons, as well as the density and atomic number, which affect the absorption efficiency. The thickness and structure of the phosphor layer determine the extent to which the visible light can laterally spread before reaching the active matrix, which affects the spatial resolution of the x-ray image.

The active matrix is an array of elements, each consisting of a sensing/storage component and a switching component. The semiconductor devices in the array are fabricated from hydrogenated amorphous silicon. In indirect detectors, the sensing component is a photodiode that detects the visible light emitted by the phosphor. A typical photodiode design is a 1–2 μm thick layer of intrinsic amorphous silicon sandwiched between thin layers of heavily-doped n-type and p-type silicon (PIN junction). A PIN diode has a larger depletion region than a PN diode and when a visible light photon is absorbed in the intrinsic layer, an electron-hole pair is generated and the electric field of the depletion region sweeps the electrons into the n^+ layer and the holes into the p^+ layer. The generated charges are stored in the junction capacitance of the diode. The photodiode may be operated under zero bias or under reverse bias. The latter increases the width of the depletion region, reduces the junction capacitance, reduces response time, and increases the dark current and noise. The switching component of an active-matrix element can be a diode or a thin-film transistor (TFT) controlled by a bias voltage. When in the *on* state, the switch connects the sensing component to the peripheral electronics but there can also be small leakage currents in the *off* state. The active matrix also contains metallic lines for switch control, for biasing the semiconductor devices, and for signal transfer from the sensing/storage elements to the peripheral electronics. The geometric fill factor is the fraction of the element area that is sensitive. Decreasing element size reduces the fill factor due to the fixed area requirements of the switching components and metallic lines.

When the switching component of an active matrix element is closed, charge is transferred from the sensing/storage component to a charge amplifier, which is primarily an op-amp with a capacitor in the negative feedback loop. The output voltage of the charge amplifier can be further amplified before arriving at the input of an analog-to-digital converter (ADC) via a shift register or a multiplexer. The output of an ADC is an integer value with a certain bit-depth. In subsequent chapters, least significant bits (LSB) will be used as

the units for dimensionless ADC quantities.

Figure 2.11: Schematic diagram of an active-matrix array and the peripheral electronics for readout. Drawn after Fig. 4.3 in Ref. [196].

Figure 2.11 is a schematic diagram of the active matrix and peripheral electronics. Active-matrix arrays are configured so that all the switches in the same row of elements are connected to the same switch control line (one switch line per row) and the outputs of the sensing/storage components in the same column are connected to the same charge amplifier via a common signal line (one signal line and charge amplifier per column). The array is first initialized and the switches are held in their *off* state while signal current from x-ray exposure (and dark current) is integrated on the capacitance of the sensing/storage elements. After exposure, the array is read out row-by-row, that is, during read out the switches in the first row are closed while the switches in other rows remain open, causing

charge to transfer from the first row of elements down the signal lines to the charge amplifiers, and then sequentially all of the remaining rows are read out. While a particular row is being read out, the other rows remain sensitive to radiation. There is a finite time required to discharge the sensing element capacitance and transfer signal to the charge amplifier, which limits the frame rate of the detector. Readout of an element also serves to initialize the element for subsequent exposure.

Due to variability inherent in the fabrication of the x-ray detection medium, the active matrix, and the peripheral electronics, the detector elements will have individual dark currents, sensitivities, and linearities. There will be some correlation in these properties among elements sharing the same charge amplifier or among elements in larger regions of the array sharing the same ADC. Component failures in the detector during fabrication or after a period of operation can cause individual elements or lines of elements to be defective. Radiation damage from excessive exposure rates or from cumulative exposure progressively increases the number of defective pixels. Manufacturers recommend replacing detectors if the percentage of defective pixels reaches 3%–5% although degradation in image quality may not be apparent for up to 20% pixel loss.¹⁹⁷

For indirect detectors, frames acquired one after another can suffer from lag, that is, signal (or lack thereof) is present in detector acquisitions subsequent to the acquisition during which it was generated. Sources of lag include incomplete charge transfer from sensing/storage elements to charge amplifiers, finite decay time of optical transitions in the phosphor, and trapping and release of charges in photodiode metastable states that arise from the disordered structure of the hydrogenated amorphous silicon.¹⁹⁸ The third effect is the dominant source of image lag.

Discrete photodiodes

It is possible to use photodiodes without a phosphor and have them be directly exposed by x-ray photons. Although less efficient, the x rays can directly create electron-hole pairs in the depletion region of the PIN junction and the liberated charge is a measure of the x-ray intensity incident on the photodiode. The scatter projection imaging systems described in Chapters 3 and 5 both utilize discrete photodiodes for the direct detection of x rays.

Chapter 3

Synchrotron-based system: Methods

The first of the two x-ray coherent scatter projection imaging systems developed for this thesis was a synchrotron-based system. This was the continuation of earlier work by Karl Landheer.¹⁸¹ In this chapter, the technical details of the apparatus, how it is used to acquire image data, and the steps taken to process the data to generate an image are described. The experimental work for this system was performed at the CLS synchrotron in Saskatoon, Saskatchewan over four trips (February 2016, September 2016, June 2017, and August 2017) spanning 48–72 hours of beam time each.

3.1 Technical details

The system was configured in the SOE-1 experimental hutch of BMIT-ID, the Insertion Device beamline of BMIT at the CLS.¹⁹⁹ Electrons circulate in the 171 m circumference Storage Ring at 2.9 GeV with an average current of 250 mA just after injection. The electron current decays exponentially with a $1/e$ time of about 23 h and electrons are re-injected every eight hours. The Storage Ring has 24 bending magnets with field strength 1.354 T.

The source of the BMIT-ID beamline is a superconducting wiggler with maximum

Figure 3.1: Schematic diagram of the experimental setup for the synchrotron-based system. The slots in the collimator can be blocked to change the number of pencil beams.

$B_0 = 4.3$ T, spatial period $\lambda_u = 4.8$ cm, and $2N_{per} = 27$ pole pairs.¹⁹⁹ The maximum values for the undulator parameter and characteristic energy are $K = 19.3$ and $h\nu_c = 24$ keV, respectively. The beamline provides an x-ray beam in the energy range 20–150 keV and has a photon brightness of 3×10^{12} ph/(s mA mrad² 0.1%BW) at 20 keV, where “mA” refers to the current in the Storage Ring and “0.1%BW” means that only photons in a 0.1% bandwidth (20 eV) centred on 20 keV are considered. Without additional collimation, the beam size is 220 mm horizontal and 11 mm vertical at 55 m from the centre of the insertion device. A pair of 2 mm-thick Si(111) bent Laue crystals monochromatize the beam to a selectable energy with 10^{-3} relative energy resolution. A beam energy of 33.2 keV was used for all experiments except when testing the effect of beam energy on the acquired images (Sec. 4.2).

See Fig. 3.1 for a schematic of the experimental setup. The beam from the monochromator was segmented horizontally into either one or five beams using a custom built collimator (Fig. 3.2).¹⁸¹ The slots in the collimator are 1.5 mm wide and adjacent slots are 15.6 mm apart centre-to-centre. Computer-controlled slits built into the beamline upstream of the experimental hutch were used for vertical collimation to give approximately square-profiled pencil beams.

Figure 3.2: Photograph of the collimator used to separate the beam from the monochromator into one or more pencil beams. Two rows of slugs with width 6.3 mm, thickness in the beam direction 3.3 mm, and material composition 56% W / 44% Cu (Eagle Alloys Corporation, Talbott, Tennessee) are placed into an Al body and spaced by Al shims 1.5 mm wide. Here five slugs have been removed from one row to produce five pencil beams. Vertical collimation is computer controlled and is done upstream along the beamline.

The object was raster scanned through the beams and was sampled in vertical strips of width equal to the beam spacing. Translational motion transverse to the beams was automated by placing the object on a platform mounted to a motorized vertical positioning stage (model KHI-4S, Kohzu Precision Co. Ltd., Kawasaki, Japan), in turn mounted atop a motorized horizontal positioning stage (model XA30A-R1, Kohzu Precision Co. Ltd.). Both stages have a motion range of 150 mm, a maximum translation speed of 10 mm s^{-1} , and $1 \mu\text{m}$ half-step resolution. The repeatabilities are $\pm 1 \mu\text{m}$ and $\pm 3 \mu\text{m}$ for the horizontal and vertical stages, respectively.

A flat-panel detector (model C9252DK-14, Hamamatsu Photonics K. K., Hamamatsu, Japan) was located approximately 6 m downstream of the collimator (58 m downstream of the centre of the insertion device) to capture x rays scattered by the object. The angle subtended at the centre of the insertion device by the width of the detector was 0.2° , and so at this large source-to-detector distance the uncollimated x-ray beam was effectively parallel. The detector has 616 rows by 1216 columns of elements $200 \mu\text{m} \times 200 \mu\text{m}$ in

size. The detector outputs a 13-bit digital value for each element per frame.

After years of use, the detector's maximum total radiation dose specified by the manufacturer (800 Gy) has been exceeded by several orders of magnitude, rendering the top 325 rows unusable. Therefore, the detector was positioned so that its bottom edge was at the height of the pencil beams, an arrangement which allowed the undamaged region of the detector to capture the scatter field out to large scattering angles but only the top half of the scatter field was recorded. The detector was interfaced to a computer via a PCI-1424 frame grabber board (National Instruments, Austin, Texas) and controlled with HiPic software version 9.3 (Hamamatsu Photonics Deutschland GmbH, Herrsching, Germany). Each object was positioned so that its exit plane was 20 cm upstream of the flat-panel detector's entrance plane.

A custom manufactured 90% W / 10% Cu blocker bar (Eagle Alloys Corporation, Talbott, Tennessee) was mounted to the detector just upstream of its entrance face to stop the transmitted primary beams from reaching the detector (Fig. 3.3). These photons would otherwise damage the detector due to their high intensity and create more scatter within the detector itself, degrading the detected distribution of scattered photons from the object. The bar thickness was chosen to be 3.0 mm, for which the transmission factor for photon energy 69.5 keV, just below the tungsten K-edge, is 6×10^{-6} and at 33.2 keV is tens of orders of magnitude smaller. In order to measure the primary x rays transmitted through the object, the blocker bar has a U-shaped channel into which discrete photodiodes with sensitive area $2.65 \text{ mm} \times 2.65 \text{ mm}$ (model BPW 34, OSRAM Opto Semiconductors GmbH, Regensburg, Germany) were mounted, positioned to intercept each pencil beam. To block as little of the low angle scatter as possible, the blocker bar was designed to have height 7.0 mm, which is the minimum that can accommodate the photodiodes. The photodiodes were connected to current-to-voltage preamplifiers (SR570—Low noise current preamplifier, Stanford Research Systems, Inc., Sunnyvale, California) and the signals were sampled.

Figure 3.3: Photograph of the 90% W / 10% Cu blocker bar on the flat-panel detector with photodiodes for measuring the primary x rays. The photodiodes and their cables are tucked into the U-shaped channel to prevent scatter from them reaching the detector.

To determine the pencil beam sizes and the relative positions of their centres at the entrance plane of the flat-panel detector, a strip of radiochromic film (Gafchromic RTQA2, Ashland, Inc., Bridgewater, New Jersey) was affixed to the detector so as to intercept the beams. The film was exposed to the pencil beams until the beam profiles were visible and then was scanned at 6400 dpi. A ruler was included in the scan, placed against the edge of the scanner window, in order to convert distances in the scan from pixels to mm. Engravings on the ruler spaced 10 mm apart were 2516 ± 7 pixels apart in the scan, which is consistent with the nominal spacing of $6400 \text{ dpi} / (2.54 \text{ cm per inch}) = 2520 \text{ pixels per cm}$. Profiles of the red pixel value were plotted across the width and height of each of the rectangular beams. The edges of the beams were not perfectly sharp and the width and height were estimated as the distance between the half-minimum points of the valley in the profile that was the pencil beam. To account for error in locating the positions of

the half-minimum points, an uncertainty of 15 pixels was estimated for each beam width and height, corresponding to 0.06 mm. The widths and heights of the five beams, at the detector's entrance plane, are shown in Table 3.1.

3.2 Data acquisition

When a pencil beam of x rays impinges on an object composed of amorphous materials, the Laue rings of scattered photons around the beam arrive at the downstream flat-panel detector. A detector acquisition of the field of scattered photons is called a *scatter pattern*. A *scatter profile* is the intensity of the scatter field from a pencil beam as a function of scattering angle, θ . If multiple pencil beams are used, the resultant scatter pattern is a superposition of the Laue rings about each individual beam, and upon disentanglement each beam has its own scatter profile. For the synchrotron-based system, since the beams are parallel, the detector face can be made perpendicular to each pencil beam and the rings are circular.

Each scatter profile is associated with one x - y location on the object. In a *scatter image*, the pixel values are integrals of the scatter profiles for the locations where the object was probed by pencil beams. In order to acquire a full set of data to produce the scatter image, the object needs to be translated such that each part of the object is irradiated by a pencil

Table 3.1: Width and height, in mm, of each pencil beam for the synchrotron-based system. The beams are numbered from left to right, viewed from the source.

Beam	Width	Height
1	1.73 ± 0.06	1.74 ± 0.06
2	1.76 ± 0.06	1.74 ± 0.06
3	1.75 ± 0.06	1.75 ± 0.06
4	1.77 ± 0.06	1.71 ± 0.06
5	1.76 ± 0.06	1.65 ± 0.06

beam. An *acquisition mode* is the manner in which the exposure of the entire object is achieved.

As stated in the outline of the thesis, an objective of scatter projection imaging system development is to reduce the time required for the acquisition of a full set of data. The use of multiple beams to simultaneously probe the object was a major improvement. In this work, data acquisition while the object is being translated continuously, rather than step-and-shoot, was implemented to further reduce acquisition time. Four acquisition modes were studied with the synchrotron-based system: single-beam step-and-shoot, five-beam step-and-shoot, single-beam continuous motion, and five-beam continuous motion. All data acquisitions were automated with a LabVIEW (National Instruments, Austin, Texas) program.

Step-and-shoot acquisition is outlined in Fig. 3.4 (a). The pencil beams sample the object in either a hexagonal or rectangular array. The acquisition parameters are the number of rows of pixels in the image, the number of columns of pixels per beam, and the horizontal and vertical spacing between pixels. For a rectangular pixel array, the horizontal and vertical spacing were always set equal to one another to give square pixels. For a hexagonal pixel array, the vertical spacing was always $\sqrt{3}/2$ times the horizontal spacing and odd rows of pixels were offset horizontally from even rows by half the pixel spacing.

Pixel data are acquired row-by-row from top to bottom, with the object travelling to the left (viewed from the source) for odd rows, and to the right for even rows. Each time the stages are translated by the spacing between pixels, the object is stopped, and a scatter pattern is acquired. The detector is operated in internal timing mode for which a sequence of back-to-back 33.3 ms detector frames is acquired, which are summed together on the frame grabber board. The number of frames summed together per scatter pattern, and hence the total acquisition time per pixel, is another parameter that can be controlled. After every scatter pattern is acquired, each photodiode's preamplifier voltage is sampled 100

Figure 3.4: (a) Flow chart for step-and-shoot acquisition. (b) Flow chart for continuous motion acquisition.

times at 1.00 kHz and the synchrotron storage ring current is recorded.

Continuous motion acquisition is outlined in Fig 3.4 (b). Data are also acquired row-by-row with the object alternately travelling left and right but the acquisition parameters are the horizontal distance travelled by the stage for each row, the number of rows, and the vertical spacing between rows. The speed, v , at which the stages translate, when not accelerating to change direction, is also specified.

When the object starts to traverse its preset trajectory, a 50% duty cycle pulse train is generated to synchronize the stage motion and data acquisition. Starting at each rising edge, all of the data for one pixel are acquired within one period of the pulse train. The pulse train period or pixel acquisition time, Δt , is determined by selecting a desired horizontal spacing between pixels within a row, Δx , when the stage is travelling at the specified speed, v , and so $\Delta t = \Delta x/v$. The horizontal pixel spacing was chosen to be equal to the vertical spacing between rows for all image acquisitions.

The flat-panel detector is operated in external timing mode, for which a frame acquisition starts at each rising edge of the external pulse and the frame acquisition time (selectable between 33.3 ms and 100 ms for the Hamamatsu C9252DK-14) is equal to the period of the pulse. Multiple frames cannot be summed into a single scatter pattern in this mode. The scatter patterns were saved as 16-bit images in a Hamamatsu HIS image sequence file, streamed to the computer's hard drive in real time.

Simultaneous to the acquisition of scatter patterns, at each rising edge of the pulse train the following sequence of events occurs: the storage ring current is recorded, the photodiodes are sampled a prespecified number of times at 1.00 kHz, the instantaneous motor positions of the horizontal and vertical translation stages are recorded, and all of these data are appended to text files. This entire sequence needs to be completed within one pulse period or else the next rising edge will be ignored, causing primary data for some pixels to be missed and scatter patterns to become mismatched with the recorded

motor positions. To fit within the pulse period, it was found that the number of photodiode samples needed to be somewhat less than half of the maximum number of samples within a pulse period (e.g., ~ 35 samples at 1 kHz for an 80 ms pulse period).

3.3 Data processing and image generation

All processing of data acquired with the synchrotron-based system was performed in MATLAB (MathWorks, Natick, Massachusetts). Image generation consists of computing the pixel values, either from the photodiode voltages for primary images or from the scatter patterns for scatter images, and placing the centres of the pixels at their sampled positions. Each image is constructed as a Voronoi diagram, that is, the plane is partitioned into convex regions such that each region contains exactly one pixel centre and every point in a given region is closer to its pixel centre than to any other. Each region is assigned a grey level corresponding to the value of the pixel it contains.

For primary images, a pixel value is generated by taking the average of the photodiode voltage samples recorded for that pixel and normalizing by the storage ring current. The x-ray fluence rate on the beamline is proportional to the ring current and it is assumed that the photodiode voltages are proportional to the fluences incident on the photodiodes. The output of the photodiodes was observed to be consistent with zero when there was no x-ray input.

For scatter images, each scatter pattern is converted into one or five pixel values, depending on how many pencil beams were used, as described in the following subsections and summarized in Fig. 3.5.

Figure 3.5: Flow chart for scatter image generation.

3.3.1 Flat-field correction

As mentioned in Sec. 2.3.1, all digital flat-panel detectors have variations in element-to-element sensitivity. It is assumed that the non-defective detector elements have a linear response in the range of exposures, X , used for our experiments,

$$P_{m,n}^{meas} = S_{m,n}^{true} X + D_{m,n}, \quad (3.1)$$

where $P_{m,n}^{meas}$ is the measured output for the element in row m and column n of the flat-panel detector, $S_{m,n}^{true}$ is the sensitivity of the element in LSB per unit exposure, and $D_{m,n}$ is the offset (output in addition to the desired signal). After flat-field correction, all of the detector elements give the same output for the same incident exposure and give zero output when there are no x rays.

The offset has a dark component and a room component. The dark offset of an element is its output when there are no incident x rays and is corrected by subtracting from each scatter pattern a dark acquisition acquired with the same integration time per frame, number of summed frames, and frame rate that the scatter pattern was acquired with. The room component arises from stray radiation that is present when the x rays are on and the collimator is in place but no object is in the beam. Scatter from the collimator, the beam stop, and in the experimental hutch adds what is assumed to be a fixed-pattern signal for a particular collimator-and-upstream-attenuation configuration (*beam configuration*) that scales linearly with the storage ring current. Multiple detector acquisitions with no object were acquired, and the ring current during these acquisitions was recorded. The appropriate dark acquisition was subtracted from each room acquisition, and the difference was divided by the frame rate to give element value due to stray radiation per second of acquisition. Plotting these values against the ring current during room acquisition, the slope of the linear fit gives stray radiation, in $\text{LSB mA}^{-1} \text{s}^{-1}$, at each element. Separate linear fits were done for

Figure 3.6: (a) Non-uniform flood field consisting of rectangular beamlets for the usual flat-panel detector calibration method. (b) Photons scattering through an angle θ out of a single beamlet (pencil beam) form a Laue ring of uniform exposure at multiple elements.

each beam configuration used in our experiments. A scatter pattern is completely corrected for offset by subtracting the dark acquisition and, for each element, subtracting the product of the stray radiation per mAs, the scatter pattern acquisition time, and the ring current during acquisition.

For transmission imaging, the usual method¹⁹⁶ for correcting the element sensitivity variation is to expose the entire detector to a flood field, $X^{flood}(x,y)$, such as the 2D beam of an x-ray tube, with no object in the beam. In general, the flood field is not uniform and the spatial variation can be expressed as $X^{flood}(x,y) = \alpha(x,y) \overline{X^{flood}}$, where $\overline{X^{flood}}$ is the mean flood field exposure over the spatial extent of the detector, which can be measured or assumed equal to some constant value, and $\alpha(x,y)$ is the unitless variation factor from that constant exposure. Rather than a continuously varying distribution, it is useful to consider the flood field as composed of a discrete set of rectangular beamlets, $X_{m,n}^{flood} = \alpha_{m,n} \overline{X^{flood}}$, each having area equal to that of a detector element (Fig. 3.6 (a)). There is a one-to-one correspondence between beamlets and elements.

The output of an element when exposed to the flood field is

$$P_{m,n}^{flood} = S_{m,n}^{true} \alpha_{m,n} \overline{X^{flood}} + D_{m,n}, \quad (3.2)$$

and the sensitivity can be estimated as

$$S_{m,n}^{calc,usual} = \frac{P_{m,n}^{flood} - D_{m,n}}{\overline{X^{flood}}} = S_{m,n}^{true} \alpha_{m,n}, \quad (3.3)$$

where $P_{m,n}^{flood} - D_{m,n}$ is the offset-corrected flood-field acquisition. The corrected detector output for subsequent acquisitions can be computed as

$$P_{m,n}^{corr,usual} = \frac{P_{m,n}^{meas} - D_{m,n}}{S_{m,n}^{calc,usual}} \times \overline{S_{m,n}^{calc,usual}}, \quad (3.4)$$

where multiplication by the average of the element sensitivities, $\overline{S_{m,n}^{calc,usual}}$, returns the scatter pattern to units of LSB with the same average value as after offset correction.

As long as the detector is not translated relative to the source after the flood-field measurement, the discrepancy between $S_{m,n}^{true}$ and $S_{m,n}^{calc,usual}$ is not an issue for transmission imaging because the same flood field is used to irradiate the object. Suppose an object has a spatially varying attenuation coefficient $\mu_{m,n}$ and thickness $t_{m,n}$. The detector acquisition will have element values

$$P_{m,n}^{meas} = S_{m,n}^{true} \alpha_{m,n} \overline{X^{flood}} e^{-\mu_{m,n} t_{m,n}} + D_{m,n}, \quad (3.5)$$

and the corrected values are

$$\begin{aligned} P_{m,n}^{corr,usual} &= \frac{S_{m,n}^{true} \alpha_{m,n} \overline{X^{flood}} e^{-\mu_{m,n} t_{m,n}} + D_{m,n} - D_{m,n}}{S_{m,n}^{calc,usual}} \times \overline{S_{m,n}^{calc,usual}} \\ &= \overline{X^{flood}} e^{-\mu_{m,n} t_{m,n}} \times \overline{S_{m,n}^{calc,usual}}, \end{aligned} \quad (3.6)$$

where the only spatial variation is due to the object and not due to element sensitivity variation or beam non-uniformity.

The discrepancy between $S_{m,n}^{true}$ and $S_{m,n}^{calc,usual}$ does, however, introduce a problem for

scatter imaging, where the pencil beam that scans over the object can be considered to be a single beamlet with exposure variation factor $\alpha_{m',n'}$ (Fig. 3.6 (b)). Two different detector elements that are irradiated by photons scattered at an angle θ out of the pencil beam will have the same incident exposure (Poisson noise not considered), $F^\theta \alpha_{m',n'} \overline{X^{flood}}$, where F^θ is a factor that depends on the scattering cross section and attenuation coefficient of the material being irradiated by the pencil beam. Since the two elements have the same incident exposure, they should have the same output after correction. However, correction with the usual method gives

$$P_{m,n}^{corr,usual} = F^\theta \overline{X^{flood}} \times \frac{\overline{S^{calc,usual}} \alpha_{m',n'}}{\alpha_{m,n}}, \quad (3.7)$$

where the non-uniformity of the flood field used for calibration results in two different values for the two elements.

Flat-field correction for scatter imaging requires a uniform flood field, $\alpha_{m,n} = 1$, for which $S_{m,n}^{calc} = S_{m,n}^{true}$. A uniform 2D flood field is not achievable in practice and so a scanning method was devised to determine $S_{m,n}^{true}$. The simplest such method involves precisely translating a single common beamlet to expose each element one after another and record their output. After measuring the absolute exposure of the beamlet using an ionization chamber and correcting for any source output variation between exposures, such as that due to decreasing storage ring current, $S_{m,n}^{true}$ can be calculated for each element. At BMIT this simple scanning method was not practical because of the time requirement to expose all 616×1216 elements. Instead, the detector was mounted on another set of translation stages (same model as the object stages) and was translated in horizontal steps equal to the element width to measure the exposure variation factors of a row of beamlets using a common detector element. Then by translating the detector vertically through that row of beamlets and normalizing by the ring current (in three passes due to the restricted width

of the beam), repeating for four different exposure levels, the sensitivity of all the detector elements was extracted.

It should be noted that at BMIT-ID it is not even possible to perform the usual flat-field correction because the uncollimated beam from the monochromator only covers an area roughly 875 elements wide by 50 elements high. In other situations it may not be possible to perform the scanning method due to, for example, a lack of computer-controlled translation stages for the detector. In this case either the usual method can be used, for which elements with the same incident exposure will have outputs varying by $1/\alpha_{m,n}$, or only an offset correction can be performed, for which elements with the same incident exposure will have outputs varying by $S_{m,n}^{true}$. Which of the two methods is preferred depends on whether the spread in $1/\alpha_{m,n}$ is smaller or larger than the spread in $S_{m,n}^{true}$.

Following flat-field correction, the values of the elements of the corrected scatter pattern were scaled to the mean ring current for the entire image acquisition and then the defective detector element values were replaced by the means of their neighbours and any negative element values were set to zero.

3.3.2 X-ray fluctuation synchronization artefact correction

The x-ray intensity at BMIT-ID oscillates with a frequency of roughly 100 Hz, thought to be caused by the fast orbit correction system²⁰⁰ which operates at a similar frequency. The flat-panel detector spends 32.9 ms of each frame reading out the element values, proceeding row by row from top to bottom. Each element integrates its signal as long as its row is not being read out. There is structureless multiple scatter from the object irradiating the whole detector, and since the x rays are always on during acquisition, even during the frame readout, oscillation in the x-ray intensity gives a horizontal striping synchronization artefact with peak-to-peak variation of about 2 LSB. Especially if a single frame is used per scatter pattern, as in continuous motion acquisition, the Laue ring intensity will be high

in some scatter patterns and low in others, giving an oscillation in the pixel values in the scatter image.

To remove the striping artefact, an array of the means of the values of the 100 right-most elements in each row of the scatter pattern, where there is minimal single-scatter signal, was filtered to remove high-frequency noise and then each row's mean was subtracted from every element in that row.

3.3.3 Disentangling scatter profiles

After correcting the scatter patterns, the scatter profiles can be extracted (one scatter profile per beam). Each scatter profile is related to the coherent scatter cross section of the material probed by its associated pencil beam and in principle is a continuous function of θ . In practice only a discrete scatter profile can be extracted since a detector with discrete elements measures the scatter. For the synchrotron-based system, the pencil beams are perpendicular to the detector face and so a fixed radial distance in the detector plane from a pencil beam corresponds to a fixed θ . If a single beam is used, the scatter profile can be simply extracted by rounding to the nearest integer the radial distance of each detector element (in units of element width) and averaging the element values over rings of constant radial distance, giving average intensity as a function of discrete radial distance. The radial distance can be converted to a scattering angle by simple trigonometry.

For multiple beams, the scatter out of each pencil beam overlaps at the detector and simple averaging over rings of constant radial distance from a pencil beam does not recover the individual scatter profiles. Instead, an iterative approach was taken. The discrete scatter profiles for the multiple beams can be stacked into a single vector $\vec{f}$ whose length is equal to the length of a single scatter profile multiplied by N_{beams} , the number of beams used for the scatter pattern. The values of the detector elements can also be flattened into a single vector $\vec{g}$. The value of the i th detector element, g_i , is the sum of exactly N_{beams} scatter

profile vector elements. This sum can be written as

$$g_i = \sum_j a_{ij} f_j, \quad (3.8)$$

where $a_{ij} = 1$ if the j th scatter profile element, f_j , contributes to the i th detector element value, g_i , and 0 otherwise. The numbers a_{ij} form a matrix $\mathbf{A}$ with number of rows equal to the number of detector elements and number of columns equal to the number scatter profile vector elements. The system matrix $\mathbf{A}$ can be constructed a priori since the positions of the pencil beams in the detector plane are known. The forward operation of calculating the scatter pattern from the scatter profiles can be rewritten as

$$\vec{g} = \mathbf{A} \vec{f}. \quad (3.9)$$

Conceptually, computing the scatter profiles from the scatter pattern amounts to multiplying $\vec{g}$ by $\mathbf{A}^{-1}$, but $\mathbf{A}$ is not square so the inverse cannot be taken directly. A computation such as $\vec{f} = (\mathbf{A}^T \mathbf{A})^{-1} \mathbf{A}^T \vec{g}$ can be attempted but the system is ill-conditioned. Equation (3.9) is instead solved by iteration using the Maximum Likelihood-Expectation Maximization (MLEM) algorithm,^{179,201,202}

$$f_j^{(k+1)} = \frac{f_j^{(k)}}{\sum_i a_{ij}} \sum_i \frac{g_i}{\sum_{j'} a_{ij'} f_{j'}^{(k)}} a_{ij}, \quad (3.10)$$

where $f_j^{(k)}$ is the current iteration's estimate of the j th scatter profile element and $f_j^{(k+1)}$ is the next iteration's estimate. The starting estimate is $f_j^{(0)} = 1$. The expression was derived initially for emission tomography by modelling the values of the detector elements as independent Poisson random variables and maximizing the likelihood of obtaining the observed values with respect to the scatter profile element intensities.

The system matrix, $\mathbf{A}$, is very sparse and it is unnecessary to store all of its elements in memory when computing the scatter profiles. Instead, two maps of indices are stored in memory. The first map is a matrix with one row per detector element and N_{beams} columns. The entries for row i are the indices, j , of elements of $\vec{f}$ that contribute to g_i . This map is used for the forward calculation $\sum_{j'} a_{ij'} f_{j'}^{(k)}$ in Eq. (3.10). The second map is an array of lists. The length of the list for the j th array element is equal to $\sum_i a_{ij}$ and the list entries are the indices, i , of $\vec{g}$ to which f_j contributes. This map is used in sums over i of terms multiplied by a_{ij} in Eq. (3.10). Iteration can continue until the distance between $\vec{f}^{(k+1)}$ and $\vec{f}^{(k)}$ is smaller than some specified value. In practice, 50 iterations were performed for each scatter pattern, since by the 50th iteration the root-mean-square change between $\vec{f}^{(k+1)}$ and $\vec{f}^{(k)}$ was found to be consistently less than 0.2%, meaning the algorithm has nearly converged to a solution.

Each scatter profile is associated with one pixel in the scatter image. The profiles were converted to grey-level pixel values by integrating them over a scattering angle range chosen to maximize contrast between specific materials. Alternatively, since the scatter profile contains an ensemble of data for each pixel, other methods that display multiple values simultaneously within a pixel are possible.¹⁸⁶

The data from each beam contribute a vertical strip of pixels in the scatter image. To correct for variation in intensity between the beams and, for primary images, variation in the detection efficiency of the photodiodes, the strips were normalized across their boundaries.

3.3.4 Flat-panel detector lag correction

For continuous motion acquisition, since the scatter patterns are recorded as back-to-back detector acquisitions 33.3–100 ms in length, the scatter patterns are affected by detector lag. This lag is propagated to the generated pixel values for scatter images.

The detector can be modelled as a linear time-invariant (LTI) system with a multiexpo-

nential impulse response function (IRF),²⁰³

$$h(k) = \sum_{m=1}^M \frac{\alpha_m}{\tau_m} e^{-k/\tau_m} H(k), \quad (3.11)$$

where $H(k)$ is the Heaviside unit-step function, k is the index of the detector acquisition acquired between times $(k-1)\Delta t$ and $k\Delta t$ after the start of exposure, and M is the number of exponential terms (we used $M = 2$ since the 3rd and additional terms have relative strength less than 1%). The m th exponential term has decay constant τ_m , measured in multiples of the acquisition interval Δt so that τ_m is dimensionless, and relative strength α_m , where $\sum_{m=1}^M \alpha_m = 1$. The fastest decaying component is typically dominant. In general, α_m and τ_m can depend on the exposure level, the time per detector acquisition, and whether the exposure level rises or falls as a beam crosses a material boundary during image acquisition, meaning the system is not truly LTI.²⁰⁴

The actual recorded output of the detector including lag for acquisition k is

$$\begin{aligned} f(k) &= h(k) * g(k) \\ &= \sum_{m=1}^M \alpha_m \left(1 - e^{-1/\tau_m}\right) \left[\sum_{l=1}^{k-1} g(l) e^{-\frac{k-l}{\tau_m}} + g(k) \right], \end{aligned} \quad (3.12)$$

where $*$ denotes convolution and $g(k)$ is the ideal output if there was no lag and can be recovered with the recursive algorithm²⁰³

$$\begin{aligned} g(k) &= \frac{f(k) - \sum_{m=1}^M \beta_m e^{-1/\tau_m} S_{m,k}}{\sum_{m=1}^M \beta_m}, \\ S_{m,k} &= g(k-1) + e^{-1/\tau_m} S_{m,k-1}, \quad S_{m,1} = 0, \end{aligned} \quad (3.13)$$

where

$$\beta_m = \alpha_m \left(1 - e^{-1/\tau_m}\right).$$

The parameters α_m and τ_m can be determined by selecting an appropriate model function for $g(k)$, finding $f(k)$ via convolution, and fitting the result to the acquired data. If the beam crosses a sharp material boundary, then the x-ray exposure at an element in the detector will abruptly rise or fall from one level A_1 to another level A_2 , and

$$g(k) = A_1 + (A_2 - A_1)H(k - k_0), \quad (3.14)$$

where k_0 is a real number giving the location of the step (that is, the largest integer $< k_0$ gives the index of the last acquisition before the step and the smallest integer $\geq k_0$ gives the index of the first acquisition after the step) and,

$$f(k) = A_1 + (A_2 - A_1)H(k - k_0) \times \sum_{m=1}^M \alpha_m \left(1 + \exp \left(-\frac{k - \text{ceiling}(k_0) + 1}{\tau_m} \right) \right). \quad (3.15)$$

When performing a detector lag correction for a scatter image acquired with continuous motion, $f(k)$ is the vector of pixel values generated from the scatter patterns, in order of acquisition. Equation (3.15) was separately fit to several rising edges in $f(k)$ and the four fit parameters α_1 , α_2 , τ_1 , and τ_2 were averaged across the values determined in each individual fit. The averages were used in the recursive algorithm (Eq. (3.13)) to compute the detector-lag-corrected pixel values, $g(k)$. The parameter values were then manually adjusted around their average fit values until the lag artefact, which manifested visually as a row-to-row pixel value oscillation within uniform material regions, appeared minimized across the entire image.

3.3.5 Pixel centre position correction

For step-and-shoot acquisition, the relative positions of pixel centres can be determined simply from the vertical and horizontal pixel spacings and the trajectory of the object dur-

ing image acquisition. For continuous motion, determining the relative positions of pixel centres is more complex due to speed and direction changes of the stages during acquisition. For this reason, the instantaneous motor positions of the vertical and horizontal stages were recorded. Furthermore, there is a delay τ_{FP} , from the time the flat-panel detector is halfway through an acquisition (the point at which the object's position corresponds to the true pixel centre) and the time the stage motor positions are sampled. Likewise, there is a delay, τ_{PD} , from the time at which the first half of the photodiode samples have been collected. Since the data are collected row-by-row with the object alternately travelling to the left and to the right, leaving the delays uncorrected yields an image with pixels in odd rows out of phase with pixels in even rows. Corrections to the recorded vertical stage motor positions are unnecessary since the vertical stage is stationary for most of the image acquisition and is only translated in one direction.

To correct the recorded horizontal stage motor positions for each pixel, the following assumptions are made: (1) the horizontal stage moves at constant speed except near row ends, (2) the direction-changing acceleration of the horizontal stage near row ends has a constant magnitude (specified in the LabVIEW program to be 100 mm s^{-2}), and (3) the magnitude of the acceleration changes abruptly from 0 to 100 mm s^{-2} as the stage approaches the end of a row and back to 0 mm s^{-2} when the specified speed is reached. Furthermore, it is assumed that τ_{FP} and τ_{PD} are constants within a particular image acquisition and that the time between acquiring motor positions for sequential pixels is equal to the pixel acquisition time Δt .

With the above assumptions, the corrected horizontal stage motor position for pixel k is

$$x_{c,k} = x_{m,k} - \frac{\tau}{\Delta t}(x_{m,k+1} - x_{m,k}) + \frac{1}{2}a\tau(\Delta t + \tau), \quad (3.16)$$

where $x_{m,k}$ is the recorded motor position for pixel k , τ is either τ_{FP} , if correcting a scatter

image, or τ_{PD} , if correcting a primary image, and a is the stage acceleration at the time of acquisition of pixel k , determined for six possible cases as

$$a = \begin{cases} 0 & x_{m,k+1} = x_{m,k} \\ 0 & |x_{m,k+1} - x_{m,k}| = \Delta x \\ -100 & x_{m,k+1} > x_{m,k} \text{ and } |x_{m,k} - x_{m,k-1}| > |x_{m,k+1} - x_{m,k}| \\ +100 & x_{m,k+1} > x_{m,k} \text{ and } |x_{m,k} - x_{m,k-1}| < |x_{m,k+1} - x_{m,k}| \\ +100 & x_{m,k+1} < x_{m,k} \text{ and } |x_{m,k} - x_{m,k-1}| > |x_{m,k+1} - x_{m,k}| \\ -100 & x_{m,k+1} < x_{m,k} \text{ and } |x_{m,k} - x_{m,k-1}| < |x_{m,k+1} - x_{m,k}| \end{cases}, \quad (3.17)$$

where the accelerations are in mms^{-2} and Δx is the specified horizontal spacing between pixels. The delays τ_{FP} and τ_{PD} are not known a priori and are determined by trial-and-error as the values that properly correct the images to have no phase difference between odd and even rows.

3.4 Image quality assessment

The methods for reducing the time to acquire x-ray scatter data, use of multiple beams and continuous motion, can result in lower quality images. Use of multiple beams results in partially overlapping scatter patterns at the detector, as demonstrated schematically in Fig. 3.1 and with real acquisitions in Fig. 4.1. This can lead to ghosting artefacts in the scatter images (as in Fig. 4.2) if the scatter profiles are not accurately disentangled. For continuous motion, if a pencil beam crosses a material boundary in the object during the acquisition of a scatter pattern, then a superposition of Laue rings associated with the two materials will be measured by the detector (see Fig. 3.7), resulting in a pixel value intermediate between the two adjacent materials.

Figure 3.7: Motion blur effect for continuous motion acquisition. Top figures show an object composed of two different materials having idealized scatter patterns. The line segments indicate the portion of the object scanned by a pencil beam during a single acquisition due to object motion in the horizontal direction. Bottom figures show the resulting scatter patterns measured by the detector. (a) The beam remained within the first material during the acquisition so only the single ring scatter pattern associated with that material was recorded. (b) The beam remained within the second material during the acquisition so only the single ring scatter pattern associated with that material was recorded. (c) The beam crossed the material boundary during the acquisition so each of the rings was recorded by the detector. The relative intensity of the two rings in the superposition depends on the fraction of time that the pencil beam irradiated each of the two materials during the acquisition.

The trade-off between imaging speed and image quality was investigated by measuring the edge sharpness in images and the image contrast-detail performance for different acquisition modes and different acquisition parameters. The SNR as a function of air kerma incident on the object was investigated as an additional image quality measure for scatter images. The plastics object in Fig. 3.8 was imaged for the SNR and edge sharpness analyses and also for comparison of images acquired with different acquisition modes and beam energies.

3.4.1 Signal-to-noise ratio

The SNR of pixel values in primary and scatter images was determined for three air kerma levels, obtained by inserting different thicknesses of PMMA upstream on the beamline and

Figure 3.8: Photo of the plastics object used for beam energy and image quality analyses of the synchrotron-based system. The outlined and labelled materials and their thicknesses are A: 2.6 cm PMMA, B: 1.8 cm PMMA, C: 0.9 cm PMMA, D: 1.0 cm polycarbonate, E: 1.0 cm polystyrene, F: 1.0 cm polyethylene, G: 1.0 cm nylon, H: 1.0 cm PMMA, I: 1.0 cm water in a plastic cuvette with 0.1 cm thick walls, J: 3.2 cm nylon, K: 3.0 cm polycarbonate, L: 3.0 cm PMMA, M: 3.0 cm polyethylene, N: 3.0 cm polystyrene, and O: 1.8 cm PMMA.

measured using an ionization chamber. A small region of the plastics object including polystyrene block E in Fig. 3.8 was imaged for each air kerma level using single-beam step-and-shoot acquisition with 0.8 mm vertical and horizontal pixel spacing and 66.7 ms acquisition time per scatter pattern. The signal was computed as the mean of the values of the 189 pixels within the polystyrene block and the noise as the sample standard deviation of those values.

3.4.2 Edge sharpness

The same region of the plastics object as in the SNR analysis was imaged using single-beam step-and-shoot acquisition and continuous motion acquisition, with different combinations of pixel acquisition time, stage speed, and pixel spacing. The horizontal width of the vertical boundary between blocks C and E and the vertical width of the horizontal boundary

between blocks E and F in Fig 3.8 were examined in the scatter images. Both boundaries are expected to have some minimal width due to aperture blur caused by the finite width and height of the rectangular pencil beams. For continuous motion acquisitions, increased boundary width is expected due to motion of the object during acquisition. Since all acquisition modes acquire data row-by-row, only vertical boundaries are expected to be affected by motion blur.

An ideal vertical boundary in the image at location $x = x_0$ between two materials with pixel value A_L to the left of the boundary and pixel value A_R to the right of the boundary can be modelled by

$$E_0(x) = (A_R - A_L)H(x - x_0) + A_L, \quad (3.18)$$

where $H(x)$ is the Heaviside unit-step function. Since the beams are rectangular, the impulse response for the horizontal aperture blur of a vertical boundary can be modelled by a rectangle function,

$$h_{aperture}(x) = \frac{1}{\tau_x} \text{Rect}\left(\frac{x}{\tau_x}\right), \quad (3.19)$$

where τ_x is the width of the pencil beam. For pixel sequences acquired with the object moving at the constant specified stage speed, v , the impulse response for motion blur can also be modelled by a rectangle function,

$$h_{motion}(x) = \frac{1}{v\Delta t} \text{Rect}\left(\frac{x}{v\Delta t}\right), \quad (3.20)$$

where Δt is the acquisition time per pixel and $v\Delta t$ is the distance travelled by the object during the acquisition of a scatter pattern.

A real vertical boundary, in images acquired without continuous object motion, is given

by

$$\begin{aligned}
E(x) &= E_0(x) * h_{aperture}(x) \\
&= \frac{A_R - A_L}{W_x} \left[\left(x - x_0 + \frac{W_x}{2} \right) H \left(x - x_0 + \frac{W_x}{2} \right) \right. \\
&\quad \left. - \left(x - x_0 - \frac{W_x}{2} \right) H \left(x - x_0 - \frac{W_x}{2} \right) \right] + A_L, \tag{3.21}
\end{aligned}$$

where $W_x = \tau_x$ is the width of the boundary, taken to be the distance over which pixel values transition from A_L to A_R . Since there is no vertical motion of the object in row-by-row acquisition, a real horizontal boundary, $E(y)$, is given by an expression equivalent to Eq. (3.21) with width W_y .

A real vertical boundary, in images acquired with continuous object motion, is given by

$$\begin{aligned}
E(x) &= E_0(x) * h_{aperture}(x) * h_{motion}(x) \\
&= \frac{2(A_R - A_L)}{W_x^2 - \tilde{W}_x^2} \left[\left(x - x_0 + \frac{W_x}{2} \right)^2 H \left(x - x_0 + \frac{W_x}{2} \right) \right. \\
&\quad - \left(x - x_0 + \frac{\tilde{W}_x}{2} \right)^2 H \left(x - x_0 + \frac{\tilde{W}_x}{2} \right) \\
&\quad - \left(x - x_0 - \frac{\tilde{W}_x}{2} \right)^2 H \left(x - x_0 - \frac{\tilde{W}_x}{2} \right) \\
&\quad \left. + \left(x - x_0 - \frac{W_x}{2} \right)^2 H \left(x - x_0 - \frac{W_x}{2} \right) \right] + A_L, \tag{3.22}
\end{aligned}$$

where $W_x = \tau_x + v\Delta t$ is the width of the boundary and $\tilde{W}_x = |\tau_x - v\Delta t|$. Equation (3.22) is identical with the exchange of W_x and $\tilde{W}_x$ and so, after fitting, the larger of the two parameters was taken as the boundary width.

The continuous motion acquisitions give images with well-defined rows but no defined columns of pixels. When determining the horizontal width of the vertical boundary between blocks C and E, Eq. (3.22) [or Eq. (3.21) for step-and-shoot] was fitted to the values

of the pixels ordered from the left to the right of the boundary for each row containing the two blocks. The values for the fit parameter W_x from each fit were averaged, and the uncertainty was computed as the standard deviation of the mean of the values. When determining the vertical width of the horizontal boundary between blocks E and F, one pixel was selected to be representative of each of several rows above and below the boundary and a single fit of the horizontal-boundary equivalent of Eq. (3.21) was made to the values of the pixels.

3.4.3 Contrast-detail evaluation

For the contrast-detail analysis, a custom phantom²⁰⁵ having a nine-row-by-ten column array of PMMA discs sitting on top of a 15 mm thick PMMA background was used, each row of discs having the same thickness (and so, the same contrast) and each column having the same diameter (Fig. 3.9). The phantom was imaged using single-beam continuous motion, five-beam continuous motion, and five-beam step-and-shoot acquisition. To evaluate the subject contrast between a PMMA disc of thickness t_D and the background of thickness t_B in the contrast-detail phantom, the following definition was used:

$$C = 2 \frac{|\Phi(t_D + t_B) - \Phi(t_B)|}{\Phi(t_D + t_B) + \Phi(t_B)}, \quad (3.23)$$

where $\Phi(t)$ is the x-ray fluence, either primary or scatter, after passing through PMMA of thickness t . Similar to equation Eq. (2.24), for a fluence of monoenergetic x rays incident on the phantom, Φ_0 , the transmitted primary fluence is

$$\Phi_P(t) = \Phi_0 e^{-\mu t}, \quad (3.24)$$

Figure 3.9: Photo of the contrast-detail phantom. Rows are labelled with the thicknesses and columns are labelled with the diameters of the PMMA discs, in mm. At the left end of each row is a steel ball bearing.

and Eq. (3.23) reduces to

$$C_P = 2 \frac{1 - e^{-\mu t_D}}{1 + e^{-\mu t_D}}. \quad (3.25)$$

The mean scatter fluence at the detector, within a scattering angle range $\theta_1 \leq \theta \leq \theta_2$ with small angles ($\theta \lesssim 10^\circ$), emerging from the phantom is

$$\Phi_S(t) \approx \frac{\Phi_0 A_0 e \sigma \rho_e}{\pi d_0^2 (\theta_2^2 - \theta_1^2)} t e^{-\mu t}, \quad (3.26)$$

where

$$_e \sigma = 2\pi \int_{\theta_1}^{\theta_2} \left(\frac{d_e \sigma_{coh}(\theta)}{d\Omega} + \frac{d_e \sigma_{inc}(\theta)}{d\Omega} \right) \sin \theta d\theta \quad (3.27)$$

is the cross section per electron for scatter (coherent plus incoherent) into the specific range of scattering angles, A_0 is the cross-sectional area of the pencil beam, ρ_e is the number of electrons per unit volume of PMMA, and d_0 is the distance from the midplane of the phantom to the detector. For scatter, Eq. (3.23) reduces to

$$C_S \approx 2 \frac{|(t_D + t_B)e^{-\mu t_D} - t_B|}{(t_D + t_B)e^{-\mu t_D} + t_B}. \quad (3.28)$$

At 33.2 keV, $\mu = 0.325 \text{ cm}^{-1}$ for PMMA.¹⁹⁵

For each image of the phantom, for every disc diameter an observer decided which thickness of PMMA disc was just distinguishable from the background. Using Eqs. (3.25) and (3.28), the contrast of that disc was determined.

The minimum detectable contrast is inversely proportional to the SNR, which in turn is assumed to be proportional to some power, k , of the object-incident air kerma: $\text{SNR} \propto K_{air}^k$. If the signal is quantum-noise limited, then $k = 0.5$. The values of k for primary and scatter were determined for the synchrotron-based system in Sec. 4.5.1. To allow comparison, minimum detected contrasts for acquisitions with different object-incident air kerma were normalized to a reference object-incident air kerma, $K_{air,ref}$, by scaling by $(K_{air}/K_{air,ref})^k$. These normalized contrasts were plotted against disc diameter to generate a contrast-detail curve.

Chapter 4

Synchrotron-based system: Images and results

This chapter presents results related to the synchrotron-based coherent scatter projection imaging system. They include data processing and image generation parameters, scatter images, data acquisition times for different acquisition modes, and experimentally-determined image quality metrics.

4.1 Data processing and image generation parameters

The standard deviation of the measured detector element sensitivities, $S_{m,n}^{true}$, was 4.0% of the mean. In absolute terms, the mean sensitivity was 5.81 LSB/(nC kg⁻¹), or 171 LSB per μ Gy air kerma incident on an element. The elements were found to have a linear response, with the mean R^2 value of the linear fits to the element-value-vs-relative-exposure data equal to 0.998.

Since the detector is not a true LTI system, it is not possible to determine a single IRF (i.e. a single set of values for α_m and τ_m in Eq. (3.11)) which corrects detector lag in all

situations. However, for a particular imaging task, it is possible to find a specific IRF that adequately corrects lag. When imaging objects consisting of different plastics, for example, for which there are moderate positive or negative changes in the x-ray fluence reaching the detector as the pencil beams cross a material boundary, the optimal lag correction parameters were found to be $\tau_1 = 33$, $\tau_2 = 0.04$, $\alpha_1 = 0.14$, and $\alpha_2 = 0.86$. Here, and in general, one decay component is much faster than the other. The strength of the slow decay component, α_1 , is moderate, so the lag is not negligible. For very uniform objects the strength of the slow decay component is close to zero and the detector lag has little effect on the image.

For continuous motion image acquisitions with 80 ms acquisition time per scatter pattern and 35 photodiode samples per pixel, the delay τ_{FP} for the pixel centre position correction was determined to be 18 ms and τ_{PD} to be 20 ms. With 100 ms acquisition time per scatter pattern and only one photodiode sample per pixel, $\tau_{FP} = -25$ ms and $\tau_{PD} = 0$ ms. The delay τ_{FP} can be positive, or can be negative (if the motor positions were sampled before the detector was halfway through the acquisition), but τ_{PD} can only be positive because the LabVIEW program was set up to record motor positions after sampling the photodiodes. Due to the subjective nature of the methods for determining τ_1 , τ_2 , α_1 , α_2 (Sec. 3.3.4) and τ_{FP} , τ_{PD} (Sec. 3.3.5), it is difficult to quantify the uncertainties of these parameters.

4.2 Comparing beam energies

The effects on scatter images of using a different monoenergetic beam energy were investigated using 33.2 keV, 50 keV, and 70 keV beams to image the plastics object shown in Fig 3.8. The coherent scatter cross section (Eq. (2.19)) implicitly depends on photon energy, E , via the magnitude of the momentum transfer argument, $|\vec{x}|$ (Eq. (2.15)). Con-

Figure 4.1: Five-beam scatter pattern recorded with beams probing the same locations on the plastics object (Fig. 3.8) for beam energy (a) 33.2 keV, (b) 50 keV, and (c) 70 keV.

sequently, an increase in the energy will result in a higher probability of scatter at lower angles. Figure 4.1 shows example scatter patterns acquired at the three beam energies with five-beam continuous motion acquisition. The object was in the same position relative to the pencil beams for each scatter pattern shown.

For these image acquisitions the wiggler field was set to 2.0 T, for which the spectrum has peak intensity near 50 keV after accounting for attenuation in the upstream attenuators and in the monochromator. The energy dependence of the x-ray intensity is the reason that the element values in the scatter patterns and the pixel values in the scatter images are at different levels for images acquired at different energies. As expected, the Laue rings contract to smaller angles at higher energies. If only a small area detector is available, then to capture as much of the single scatter as possible a higher beam energy can be used. In this situation, however, the primary blocker should be made as small as possible since much of the structure in the scatter field is near the primary beams. Higher energy beams

should also be used if the object is highly attenuating or very thick.

Figure 4.2 shows the resulting scatter images for the three beam energies. To give comparable images when using different energies, the scatter profiles were integrated over the same range of $|\vec{x}|$, rather than the same range of θ . The materials appear to have mostly the same contrast. However, attenuation differences at the different energies give some variation between the images.

Figure 4.2: Five-beam continuous motion scatter images of the plastics object (Fig. 3.8) acquired with beam energy (a) 33.2 keV, (b) 50 keV, and (c) 70 keV. For all three energies, the scatter profiles were integrated over momentum transfer argument $0.701 \text{ nm}^{-1} \leq |\vec{x}| \leq 0.935 \text{ nm}^{-1}$, corresponding to $3.00^\circ \leq \theta \leq 4.00^\circ$ for 33.2 keV. These images, and subsequent scatter images in this chapter and in chapter 6, were generated as Voronoi diagrams, as described in Sec. 3.3.

4.3 Acquisition modes

Figure 4.3 shows scatter images of the plastics object in Fig. 3.8 acquired with the four different acquisition modes studied with the synchrotron-based system. Acquisition parameters were chosen to give the shortest acquisitions for each mode. Table 4.1 lists the total time to acquire all of the scatter data for each image. There is approximately a thirty-fold acquisition time reduction when changing from step-and-shoot to continuous motion acquisition.

Figure 4.3: Comparison of scatter images for different acquisition modes, scatter profiles integrated $3.00^\circ \leq \theta \leq 4.00^\circ$. (a) Single-beam step-and-shoot (slowest acquisition), (b) five-beam step-and-shoot, (c) single-beam continuous motion, and (d) five-beam continuous motion (fastest acquisition). The number of pixels in the images is (a) 7104, (b) 7040, (c) 7380, and (d) 8425.

Going from single-beam to multi-beam acquisition results in some ghosting artefacts, due to imperfect extraction of scatter profiles from the scatter pattern using the MLEM disentangling algorithm, and some stitching artefacts, due to imperfect normalization between beam strips. In the case of five-beam acquisition, there was a fivefold acquisition time re-

Table 4.1: Total time in minutes to acquire scatter data for each image in Fig. 4.3.

	Single beam	Five beam
Step-and-shoot	305.77	61.37
Continuous motion	9.84	2.25

duction compared to single-beam acquisition for step-and-shoot. Five beam continuous motion acquisition was 4.37 times faster than single-beam continuous motion acquisition. The time reduction was less than fivefold because the total time spent reversing at the ends of rows remained the same but made up a larger fraction of the total acquisition. For a horizontal arrangement of beams, it would be more efficient to acquire data column-by-column since the total number of travel direction reversals would be reduced. However, there would also be a risk of motion artefacts in medical imaging.

The air kerma incident on the object for the images in Fig. 4.3, as determined from measurements with an ionization chamber (model 31010 used with electrometer model UNIDOS weblne, PTW-Freiburg, Freiburg, Germany), are shown in Table 4.2. For step-and-shoot, these air kerma calculations assume that the x-ray beam was shuttered between detector acquisitions, which was not the case in practice. Continuous motion acquisition does not require shuttering. For the plastics object, which has thickness 0.9–3.2 cm, object-incident air kerma less than about 1 mGy resulted in scatter fluence being at the lower detection limit of the detector.

Table 4.2: Average object-incident air kerma (mGy) for each image in Fig. 4.3.

	Single beam	Five beam
Step-and-shoot	13.7 ± 2.0	17.1 ± 2.5
Continuous motion	8.4 ± 1.2	8.2 ± 1.2

Figure 4.4: Source-facing side (left) and detector facing-side, horizontally flipped (right) of a 3.5 cm thick pork sample (two 1.75 cm thick cuts, one behind the other) placed in a plastic container filled with PBS.

4.4 Biological images

A 3.5 cm thick pork sample (two 1.75 cm thick cuts, one behind the other) in a slab-sided PMMA container filled with phosphate-buffered saline (PBS) (Fig. 4.4) was imaged using single-beam step-and-shoot acquisition. Figure 4.5 shows the primary and scatter images of the acquisition, for which pixels were acquired in a hexagonal array with 1.68 mm horizontal spacing. Scatter profiles were integrated $3.5^\circ \leq \theta \leq 5.6^\circ$.

The primary image is excellent for identifying bone but has poorer fat-muscle contrast. Indicated in each image are seven-pixel regions of fat and muscle whose pixel values were used to compute fat-muscle contrast as

$$C = 2 \frac{|\bar{p}_1 - \bar{p}_2|}{\bar{p}_1 + \bar{p}_2}, \quad (4.1)$$

where $\bar{p}_1$ and $\bar{p}_2$ are the means of pixel values in regions of interest in the first and second materials, respectively. The fat-muscle contrast is 0.088 ± 0.013 for the primary image and 0.65 ± 0.06 for the single-beam scatter image.

Figure 4.5: Single-beam (a) primary and (b) scatter images of the pork object in Fig. 4.4. Outlined in each image are seven-pixel regions of muscle and fat.

4.5 Image quality assessment

4.5.1 Signal-to-noise ratio

If the image data are quantum-noise limited, the pixel SNR will vary as the square root of the object-incident air kerma for both primary and scatter images. If the exposure is too low, the readout noise of the detector, which is signal-size independent, will become comparable to the quantum noise and the SNR will vary faster than the square root of the air kerma, approaching a linear relationship when the data are readout-noise limited.

Figure 4.6 plots pixel SNR as a function of air kerma, K_{air} , incident on the object for primary and scatter images. The error bars for the SNR are too small to be visible. Power laws were fit to each set of data giving $SNR \propto K_{air}^{0.39 \pm 0.03}$ for primary and $SNR \propto K_{air}^{0.72 \pm 0.05}$ for scatter. The noise varied as $K_{air}^{0.57 \pm 0.04}$ for primary and as $K_{air}^{0.18 \pm 0.03}$ for scatter.

Figure 4.6: SNR calculated using the pixels within a region of polystyrene block E (Fig. 3.8) for different air kerma levels at the object, for both primary and scatter images. Solid lines are power-law fits to the data. Triangles and circles are SNR estimates for primary and scatter, respectively, had ideal photon-counting detectors been used.

The exponent for the scatter data fit is greater than 0.5 but less than 1.0, indicating an intermediate SNR regime between quantum- and readout-noise limited. The readout noise for the Hamamatsu C9252DK-14 detector for a two-frame scatter pattern is approximately 3.6 LSB, according to the manufacturer's manual. At the lowest air kerma level used, the element values in the detector region integrated to generate pixel values were less than 7 LSB, so the image data are nearly readout-noise limited. The exponent for the primary data fit is less than 0.5, suggesting the data are quantum-noise limited. This is expected since the air kerma at the photodiode was very high (greater than 0.7 mGy).

Also included in Fig. 4.6 are estimates of the maximum SNR for the primary and scatter pixel values if ideal photon-counting detectors were used. The number of primary and coherently scattered quanta behind the object was estimated from the object-incident air kerma, and it was assumed that all photons were counted. The primary and scatter measurements are about one and two orders of magnitude lower than ideal, respectively. Two factors which lower the measured pixel SNRs are geometry (e.g., the primary beam not aligned with the photodiode, less than half of the scatter Laue ring intercepted by the detector) and quantum efficiency. Furthermore, the scattered photons are spread over a much larger area than the primary photons and the scatter energy deposited per detector element is 0–10 LSB, which is at or near the detection limit, significantly reducing the SNR.

4.5.2 Edge sharpness

One single-beam step-and-shoot, five single-beam continuous motion, and five five-beam continuous motion acquisitions were made for a region of the plastics object around block E in Fig. 3.8. For the single-beam acquisition, the left-most pencil beam (beam 1 in Table 3.1) crossed the boundaries of interest. For the five-beam acquisitions, it was the middle beam (beam 3 in Table 3.1). The acquisition parameters for each image as well as the computed widths for vertical and horizontal boundaries are listed in Table 4.3. Figure 4.7 shows

Table 4.3: Horizontal and vertical boundary widths for images with different acquisition parameters. Image 1 is a single-beam step-and-shoot acquisition. Images 2–6 are single-beam continuous motion and images 7–11 are five-beam continuous motion acquisitions. For all acquisitions, the vertical spacing was the same as the horizontal spacing. For Image 1, the pixel spacing was 0.8 mm and for all others it was $v\Delta t$.

Image #	Δt (ms)	v (mm/s)	$v\Delta t$ (mm)	W_x (mm)	W_y (mm)
1	66.7	0.0	0.0	1.713 ± 0.005	1.775 ± 0.020
2	100.0	10.0	1.0	2.82 ± 0.10	N/A
3	100.0	7.0	0.7	2.40 ± 0.07	1.86 ± 0.11
4	100.0	5.0	0.5	2.16 ± 0.05	1.80 ± 0.13
5	100.0	3.0	0.3	2.04 ± 0.04	1.69 ± 0.06
6	50.0	10.0	0.5	2.23 ± 0.07	1.74 ± 0.16
7	100.0	10.0	1.0	2.98 ± 0.17	N/A
8	100.0	7.0	0.7	2.50 ± 0.08	1.70 ± 0.13
9	100.0	5.0	0.5	2.27 ± 0.04	1.63 ± 0.15
10	100.0	3.0	0.3	2.05 ± 0.04	1.76 ± 0.08
11	50.0	10.0	0.5	2.22 ± 0.06	1.48 ± 0.14

Figure 4.7: Pixel value vs pixel centre horizontal coordinate for a portion of one row in Image 5 of Table 4.3 in the vicinity of the vertical boundary between blocks C and E in Fig. 3.8. Equation (3.22) was fitted to the data with fit parameters shown in the figure. The horizontal width for this row is 2.1 ± 0.3 mm.

one example of the transition in pixel values across a vertical boundary along with a fit of Eq. (3.22).

The blurred boundary models (Eqs. (3.21) and (3.22)) give $W_x = \tau_x + v\Delta t$. Hence, plot-

Figure 4.8: Horizontal widths for the vertical boundary between blocks C and E in Fig. 3.8 vs $v\Delta t$ for the continuous motion scatter images in Table 4.3. Widths for single-beam and five-beam images are plotted separately, each with a linear fit.

ting W_x against $v\Delta t$ for continuous motion images (Fig. 4.8) should give a line with unity slope and intercept equal to the width of the pencil beam crossing the boundary of interest. The pencil beam width (via film, Sec. 3.1) was 1.73 ± 0.06 mm for the single-beam acquisitions and 1.75 ± 0.06 mm for the five-beam acquisitions. The single-beam fit has slope 1.02 ± 0.13 mm and intercept 1.70 ± 0.07 mm. The five-beam fit has slope 1.18 ± 0.17 mm and intercept 1.68 ± 0.08 mm. Both intercepts are within uncertainty of the measured beam widths, and so is the computed horizontal width for the single-beam step-and-shoot acquisition. Therefore, it is reasonable to model W_x as being proportional to $v\Delta t$ and having a lower limit equal to the beam width τ_x . The motion blur increases the vertical boundary width by up to $v\Delta t = 1.0$ mm, which amounts to a 50%–100% increase if the pencil beam width is in the range of 1–2 mm. This has little effect on the images presented in this chapter but could be important if looking for small features such as microcalcifications.

The mean vertical width for all images in Table 4.3 is 1.72 ± 0.04 mm, within uncertainty of the measured beam heights (1.74 ± 0.06 mm for single-beam acquisitions and 1.75 ± 0.06 mm for five-beam acquisitions). This result agrees with the expectation that horizontal boundaries are not affected by motion blur for our acquisition modes and that the vertical widths of horizontal boundaries should be equal to the height of the rectangular beams.

For continuous motion acquisition, $v\Delta t$ is equal to the horizontal spacing between pixels, Δx . Only two of the three parameters can be selected independently, within the limits of the detector and the translation stages. A smaller Δx gives better spatial resolution and less boundary blur, a larger v gives a shorter overall image acquisition time, and a larger Δt gives higher SNR. Therefore, a compromise must be made between the three parameters when acquiring an image.

4.5.3 Contrast-detail evaluation

Table 4.4 lists the acquisition parameters and acquisition modes for the three pairs of images of the contrast-detail phantom (Fig. 3.9) acquired for the contrast-detail analysis. The relative air kerma rate incident on the phantom depends on the thickness of PMMA attenuator inserted upstream on the beamline and on the storage ring current during acquisition. Figure 4.9 is an example primary image of the phantom.

Figure 4.10 shows the air-kerma-normalized contrast-detail curves using $k = 0.39$ for primary images and $k = 0.72$ for scatter images. On these plots, better-performing systems will have curves closer to the lower-left corner, since small features in the image can be detected even if they have lower contrast with the background. For primary images, air kerma is proportional to the product of the air kerma rate at the phantom and the number of photodiode samples. For scatter images, air kerma is proportional to the product of the air kerma rate and Δt . The overall object-incident air kerma relative to the single-beam

Table 4.4: Acquisition parameters and object-incident air kerma, relative to the single-beam continuous motion image, for the primary and scatter images used for the contrast-detail analysis. Factored into the relative air kerma is an overlap factor, the number of times a region of the phantom was exposed by a pencil beam (4 for step-and-shoot, 2 for continuous motion for these images). The acquisition modes are 1B-CM: single-beam continuous motion, 5B-CM: five-beam continuous motion, and 5B-SS: five-beam step-and-shoot.

Acquisition mode	Δt (ms)	Δx (mm)	# photodiode samples	Relative air kerma rate	Relative primary air kerma	Relative scatter air kerma
1B-CM	80.0	0.8	30	1.000	1.000	1.000
5B-CM	80.0	0.8	35	0.136	0.159	0.136
5B-SS	66.7	0.8	100	0.127	0.842	0.212

Figure 4.9: Five-beam step-and-shoot primary image of the contrast-detail phantom. Rows are labelled with the thicknesses, and columns are labelled with the diameters of the PMMA discs, both in mm. At the left end of each row is a steel ball bearing.

Figure 4.10: Air-kerma-normalized contrast-detail curves for (a) primary and (b) scatter images in Table 4.4.

continuous motion image is also shown in Table 4.4. The curves in Fig. 4.10 overlap, indicating that contrast-detail performance is independent of the acquisition mode. The one exception is the scatter image curve for five-beam step-and-shoot, for which it was difficult to distinguish any discs from the background because the experiment was done with too low an exposure and could not be repeated. The five-beam step-and-shoot curve is included for completeness.

4.6 Synchrotron-based system summary

A coherent x-ray scatter projection imaging system using multiple beams and a continuously moving object was implemented on the BMIT-ID beamline at the CLS synchrotron. Several corrections had to be performed when generating scatter pixel values from the raw scatter patterns.

A goal of the implementation was to make image acquisition as short as possible. Con-

tinuous motion acquisition was determined to be about thirty times faster than step-and-shoot acquisition and using multiple beams reduces acquisition time by a factor nearly equal to the number of beams.

The trade-off between imaging speed and image quality was investigated by measuring the edge sharpness in images and the contrast-detail performance for different acquisition modes. Use of continuous motion increases the widths of edges in the scatter image by the product of the object translation speed and the acquisition time per pixel. Contrast-detail performance was found to be dependent on object-incident air kerma but not on acquisition mode.

SNR measurements for different object-incident air kerma levels indicate that, in addition to quantum noise, the readout noise of the flat-panel detector is a limitation. A detector with lower readout noise and with greater sensitivity could improve scatter image SNR. The scatter pattern usually does not need to be acquired at high resolution and so a detector with larger elements or one capable of binning elements to reduce additive readout noise would be useful.

Chapter 5

X-ray tube-based system: Methods

A synchrotron light source can provide high intensity monoenergetic x rays and so is an excellent facility for the purpose of developing an x-ray coherent scatter projection imaging system. It is very expensive and complicated, however, to construct and commission a synchrotron facility and consequently it has limited accessibility, making it impractical for day-to-day imaging tasks. Therefore, in a move toward practical application of coherent scatter projection imaging, a second system was developed using the main components of medical and industrial radiological imaging: a rotating anode x-ray tube, a high voltage generator, and an energy-integrating flat-panel x-ray detector.

Implementing an x-ray tube-based coherent scatter imaging system is made difficult due to the low photon flux (compared to a synchrotron light source), which combined with the fact that the scattered photons are spread over a large area at the detector plane means that the information-rich scatter can be undetectable if the detector is not sufficiently sensitive. In addition, the polychromatic beam blurs scatter profiles, reducing contrast. The optimization of and experimentation with the synchrotron-based system informed the design and development of the x-ray tube-based system. In this chapter, the design choices and technical details of the x-ray tube-based system are presented and the scatter data

Figure 5.1: Schematic diagram of the experimental setup for the x-ray tube-based system.

acquisition and processing procedures are described.

5.1 Design considerations

The basic geometry of our x-ray tube-based coherent scatter projection imaging system is shown in Fig. 5.1. X rays are provided by a Machlett Dynamax rotating tungsten anode tube (69MB insert in a 62R housing) which has a target angle of 12° , nominal large focal spot size 1.2 mm, and is rated for 25–125 kVp. The maximum thermal energy that may be stored in the anode is 220 kJ and the maximum cooling rate is 925 W. The x-ray tube is powered by a Picker GX550 single-phase full-wave rectified generator and the anode rotor is driven at a rotation speed of 3000 rpm.

To determine the x-ray tube output, a 15 cm^3 pancake diagnostic chamber (PRM model D-15, Precision Radiation Measurement Inc., Nashville, Tennessee) was placed in the uncollimated beam 35 cm from the focal spot. The integrated ionization current was measured at 110 kVp with 1.0 mm of Al filtration for twelve different mAs settings ranging from 1.7–50.0 mAs and the response was found to be linear over the entire range. Converting the

integrated current to air kerma and fitting a line, the air kerma yield of the x-ray tube was measured to be about $530 \mu\text{Gy mAs}^{-1}$ at 110 kVp with 1 mm of Al filtration, 35 cm from the focal spot.

X rays are detected using a Perkin Elmer XRD 0840 AN13 flat-panel detector (Varex Imaging Corporation, Salt Lake City, Utah), which has 512×512 elements with $400 \mu\text{m}$ pixel pitch. The detector has a $\text{Gd}_2\text{O}_2\text{S}$ scintillator, 16 bit digital output per element, and a maximum frame rate of 7.5 frames/s when externally triggered. The detector has variable gain, which is changed by internally switching the feedback capacitors of the preamplifiers. For the highest gain setting (0.5 pF), the sensitivity of the detector elements was measured to be approximately 15 000 LSB per μGy of incident air kerma at 110 kVp with 1 mm of Al filtration. The detector is interfaced to a computer using a Perkin Elmer XRD-FG frame grabber board.

For a 10 mAs exposure, the air kerma at 35 cm from the focal spot can be as high as about 5 mGy. While this is far lower than what the BMIT beamline is capable of producing, for scatter image acquisitions using the synchrotron-based system, the x-ray beam intensity and acquisition parameters were chosen so that the object-incident air kerma was in the range of 1–30 mGy, typically around 5 mGy. Furthermore, the Perkin Elmer detector is nearly 100 times more sensitive than the Hamamatsu detector used at BMIT, which had an estimated sensitivity of 170 LSB per μGy of incident air kerma. This suggested that a coherent scatter imaging system built using the Machlett Dynamax x-ray tube and the Perkin Elmer flat-panel detector would produce scatter patterns well above the noise floor of the detector for reasonable acquisition times.

As with the synchrotron-based system, in addition to the x-ray source and detector, the other key components are the collimators between the source and the object, and the object translation stages. Unlike the beam from the monochromator at a synchrotron light source, an x-ray tube beam is two-dimensional, and the cone of radiation can be formed by the

collimators into a two-dimensional array of pencil beams, allowing for more scatter image pixels to be acquired simultaneously compared to the synchrotron-based system. Since the x-ray tube can only output a beam of sufficient intensity in discrete, intermittent exposures, the acquisition of data for the entire scatter image needs to be done in a step-and-shoot manner. The flat-panel detector acquires the overlapping scatter signals from the multiple beams irradiating the object. The same detector is used for the simultaneous measurement of the transmitted primary beams. Attenuator discs are placed on the entrance face of the flat-panel detector to reduce the intensity of the transmitted beams to the same level as the typical scatter intensity at a detector element. A design decision was to make the system lead-free, with all collimators and attenuators made from an alloy 90% tungsten by weight.

The design parameters for this geometry are the source-to-object distance, s , the object-to-detector distance, d , the beam size at the object, the arrangement and spacing of the pencil beams at the object, the tube voltage (due to the use of attenuators for primary beam measurement), and the maximum size and weight of the object. The system was designed to be capable of imaging objects with mass up to 1 kg, width and height up to 10 cm, and thickness up to 10 cm, but it is expected that objects imaged with the system will typically have thickness less than 6 cm. Such a system would be useful for imaging small packages and samples of soft tissue and bone.

Ray-tracing simulations were performed using Python to calculate the coherent and incoherent single scatter intensity at each detector element as well as the resulting scatter profiles, including spectral and thickness blur, for different combinations of the design parameters. The inputs to the simulations were the measured properties of the x-ray tube and flat-panel detector, x-ray tube spectrum data from Birch et al.,¹⁹³ coherent scatter form factor and incoherent scattering function data from King et al.,¹⁷³ and attenuation coefficient data from the NIST XCOM database.¹⁹⁵ Poisson noise was included in the scatter intensity measured by the detector elements. To validate the simulations, scatter profiles were mea-

Figure 5.2: Comparison of simulated and measured scatter profiles for polyethylene (left) and polystyrene (right). Source-to-object distance: 35 cm, object-to-detector distance: 12.5 cm, object thickness: 1 cm, beam diameter: 3 mm, tube voltage: 110 kV. The peak regions mostly agree after scaling the measured scatter profiles to have the same maximum value as the simulated profiles.

sured for two materials for one set of simulation parameters (Fig. 5.2). It was found that the shapes of the peaks of the simulated scatter profiles match the experimental measurements after applying a scale factor.

For fixed d , the intensity of the primary beam at the object and therefore the scatter intensity at the detector should decrease as $1/s^2$, and this behaviour is observed in the computed scatter profiles (see Fig. 5.3 and Table 5.1). Distance s should be minimized to increase intensity but should not be less than about 25 cm due to the practical requirement of space for installing collimators between the source and the object. Furthermore, increasing the ratio d/s increases the magnification of the finite focal spot and consequently the blurring of scatter profiles. A nominal distance of $s = 35$ cm from the focal spot to the midplane of the object was selected.

Increasing the diameter of the pencil beams provides more primary photons to scatter from the object. However, smaller beams are desired to reduce the blurring of the scatter

Figure 5.3: Simulated scatter profiles for a 1 cm thick polyethylene block with fixed $d = 10$ cm for different source-to-object midplane distances, s , at 110 kV.

Table 5.1: Maxima of peaks of scatter profiles in Fig. 5.3.

s (cm)	Peak max. (LSB)	Peak max. $\times (s/25)^2$
25	1640.7	1640.7
35	836.5	1639.6
45	505.7	1638.3
55	338.1	1636.6
65	242.2	1637.4
75	181.8	1635.8
85	141.3	1634.0
95	113.2	1634.1
105	92.6	1632.8

profiles and to improve the spatial resolution of the projection image. During the design phase of the x-ray tube-based system, acquisitions with balanced filters were considered. That is, two scatter patterns would be acquired, one with an Er filter and the second with a Tm filter, and the Er-filtered scatter pattern would be subtracted from the Tm-filtered scatter pattern to yield a pseudomonoenergetic acquisition. Although the balanced filters were ultimately not used for the experimental work in this thesis, in part due to their dose inefficiency, their potential future use influenced the beam size as follows.

Use of balanced filters is only worthwhile if the scatter profile resolution gain is not lost to other blurring processes. Two sets of scatter profiles were simulated using a 110 kV beam, the first with an Er filter and the second with a Tm filter. It was found that a diameter of 3 mm is approximately the largest beam size for which the two major peaks in the differential coherent scatter cross section for polyethylene were resolvable after taking the difference of the profiles acquired with Tm and Er, and so this was chosen as the beam diameter for the system. The beam diameter at the object plays a role in determining the required diameter of the primary attenuator discs. If the diameter of the attenuators is too large then they will also block the low-angle scatter, depending on how far the detector is from the point of scatter.

Decreasing d increases the solid angle subtended at the scattering centre by each detector element, increasing the measured scatter intensity. Moving the detector from 10 cm to 5 cm behind the object increases the peak scatter intensity by a factor of about 3.5 (Fig. 5.4). However, having an element intercept photons from a larger range of scattering angles reduces the angular resolution of the measured scatter profiles. Furthermore, if insufficient distance is provided for the scatter to spread out, many scatter photons can fall on detector elements that are shielded by the primary attenuators. As d increases and the scattering angles spread out over more detector elements, the profile resolution improves somewhat but the drop in scatter intensity is much more dramatic. It was decided to use the minimum

Figure 5.4: Simulated scatter profiles for a 1 cm thick polyethylene block with fixed $s = 35$ cm for different object midplane-to-detector distances, d , at 110 kV. The scatter profiles are plotted as a function of radial distance in the detector plane from the centre of the primary beam.

distance that allows for the bulk of the Laue rings to clear the edges of the primary attenuator. A nominal distance of $d = 12.5$ cm from the midplane of the object to the detector was selected, which was found to be compatible with the necessary primary attenuator diameter of 7.0 mm for this distance. This attenuator diameter accounts for the increase in the primary beam size from the object to the detector, the finite size of the focal spot, and a margin of adjustment for positioning.

X-ray output increases greater than the second power of the kVp, taking filtration into account. The nominal tube voltage of the system was selected to be 110 kV, as it is within the limits of the x-ray tube and provides a high flux of penetrating x rays. Other tube voltages were also investigated after the system was built, as discussed in Sec. 5.4.1. For 110 kV with 1 mm of Al, 90% W / 10% Cu primary attenuators 1.5 mm thick give transmitted x-ray intensities of the same magnitude as the typical scatter intensity at a detector element.

Simulations showed that at 110 kV with 1 mm Al, the maximum scatter intensity for

Figure 5.5: Simulated scatter profiles for varying thickness of a polyethylene block with fixed $s = 35$ cm and $d = 10$ cm. The x-ray beam spectrum is 110 kV with 1 mm Al.

Figure 5.6: Maximum values of the scatter profiles in Fig. 5.5 vs thickness of the polyethylene block.

polyethylene is for a thickness of about 6 cm (Fig. 5.5), approximately equal to the inverse of the linear attenuation coefficient. For clarity, the maximum values of the scatter profiles in Fig. 5.5 are shown in Fig. 5.6 as a function of polyethylene thickness. For thicker objects, the intensity drops off due to increased attenuation. The blurring of scatter patterns, however, progressively increases with thickness since photons of the same energy scattered at the same angle at different positions along the pencil beam in the object will land on different elements of the detector. It is expected that the system will be typically used to image objects with thicknesses in the range of 1–6 cm.

To make faster image acquisitions, up to fifteen beams in a rectangular arrangement of three rows and five columns can probe the object simultaneously. The middle beam is along the system axis (the line through the focal spot and perpendicular to the detector face). The columns of beams are spaced by 2.0 cm at 35 cm from the source so that a 10 cm wide object placed at the nominal source-to-object distance can have its entire width sampled without overlap. The rows are also spaced by 2.0 cm. Due to the anode heel effect, different rows of beams have different intensities. Arrays with fewer beams can be used by blocking some of the beams.

5.2 Technical details

Figure 5.7 shows the positions of system components along the system axis. The collimation between the source and object consists of three 90% W / 10% Cu plates, a 90% W / 10% Cu diaphragm in the tube port, and a 90% W / 7% Ni / 3% Fe diaphragm insert, each custom-made (Eagle Alloys Corporation, Talbott, Tennessee). Figure 5.8 is a drawing of plate 3. Each of the three plates is 2.0 mm thick and is square with side length 16.0 cm. The middle beam of the 15-beam array passes through the centre of each plate. Since the x-ray beam is diverging, the fifteen holes in each plate are focused on the focal spot and the plates

Figure 5.7: Positions (cm) of the x-ray tube-based system components along the system axis, relative to the focal spot. The tube is inclined so that the beam's central axis is 6.1° above the system axis.

have fixed positions along the z -axis. Plate 3 farthest from the source gives the pencil beams their final size. Its midplane is 23.3 cm from the source, the holes have 2.0 mm diameters, and are spaced by 13.3 mm at the midplane. Plate 1 nearest to the source restricts off-focal radiation. Its midplane is 8.0 cm from the source, the holes have 0.70 mm diameters, and are spaced by 4.56 mm at the midplane. Plate 2 blocks primary x rays that would otherwise pass through unmatched holes in plates 1 and 3. Its midplane is 17.5 cm from the source, the holes have 2.0 mm diameters, and are spaced by 10.0 mm at the midplane. Blocker bars (90% W / 10% Cu, Midwest Tungsten Service, Willowbrook, Illinois) 5 mm thick can be attached to the rear face of plate 2 to selectively block x rays from passing through some of the holes in the plate.

Figure 5.9 is a photograph of the complete system. Each plate is mounted to a Velmex Unislide A15 manual translation stage (Velmex, Inc., Bloomfield, New York) for fine positioning in the x -direction. The stages have 3.8 cm of travel and the lead screw advance is 1.00 mm per turn. The stages are mounted to optical posts held by VPT-2 translating optical post holders (Newport Corporation, Irvine, California) that have 7.1 mm of vertical adjustment and allow for fine positioning in the y -direction. The post holders stand on

Figure 5.8: Drawing of collimator plate 3 of the x-ray tube-based system, showing the arrangement of the holes. The holes around the edge of the plate are for mounting purposes. The dashed lines are the fifteen diverging pencil beams. The fifteen central holes are angled toward the focal spot and have diameter 2.0 mm. The plate is 2.0 mm thick and 16.0 cm on each side.

optical rail carriers that can be moved along an optical rail for coarse positioning in the z -direction. The pitch (rotation about the x -axis) and roll (z -axis) are fixed. The yaw can be coarsely adjusted by rotating the optical post within the holder. The degrees of freedom for each plate are adjusted independently of one another and they must be carefully aligned with each other and with the source.

The tube is mounted vertically (anode below cathode) and inclined 6.1° (anode end toward the object) to give an effective target angle of 5.9° , reducing the apparent size of the focal spot while maintaining sufficient intensity in the lower row of pencil beams. The diaphragm (Fig. 5.10) is fitted in the tube housing port and has a conical shape extending into the port. The source-facing side of the diaphragm is approximately 3.4 cm from the centre of the focal spot and the 18 mm diameter circular hole is parallel to the tube port.

Figure 5.9: Photograph of the rotating-anode x-ray tube coherent scatter projection imaging system.

Figure 5.10: Left: drawing of diaphragm and diaphragm insert viewed from the source side. The small post on the insert fits into the matching hole in the diaphragm to fix the orientation of the insert. Right: photograph of the diaphragm and insert inside the tube port, viewed from the object side. The nylon screws in the plate securing the diaphragm fix the orientation of the diaphragm with respect to the tube housing.

A thin aluminum plate secures the diaphragm in the tube port. Holes in the plate are aligned with holes on the object side of the diaphragm and screws fix the orientation of the diaphragm with respect to the tube housing. The diaphragm insert is a 25 mm diameter, 7.5 mm thick disc with two holes. The larger, rectangular hole restricts the x-ray beam to a rectangular field illuminating the 15 holes in collimator plate 1. The hole is diverging and angled in the disc to account for the tube incline and to produce a beam centred on the system axis. The smaller hole is circular with a 2.0 mm diameter and directs a beam to collimator plate 1, 1.0 cm above the top row of holes, where a beam monitor is placed. The centre of the insert is approximately 4.0 cm downstream of the focal spot and it is held in place by gravity. A small post on the source side of the insert slides into a hole in the diaphragm to fix its orientation with respect to the tube housing. The magnetic permeability of the diaphragm insert allows for easy removal from the diaphragm using a strong permanent magnet. The shape of the field from the x-ray tube can be changed easily by swapping inserts.

A photodiode with sensitive area 6.55 mm^2 (model BPW 34, OSRAM Opto Semiconductors GmbH, Regensburg, Germany) is placed on the source side of collimator plate 1 and in the path of the smaller, circular beam from the diaphragm insert. This photodiode monitors the variation in the x-ray tube output from one exposure to another. A 0.6 mm sheet of copper is placed in front of the photodiode to reduce the x-ray intensity and prevent signal saturation. The photodiode is zero-biased with the anode grounded and the cathode connected to the input of a charge amplifier built using an LMC6482 dual op-amp (Texas Instruments Inc., Dallas, Texas).

Two Velmex XSlide motorized linear translation stages (Velmex, Inc., Bloomfield, New York) are configured to translate the object in the *x*- and *y*-directions. The stages have 15 cm of travel and the lead screw advance is 2.00 mm per turn. Vexta PK245-01BA stepper motors (Oriental Motor Co., Ltd., Tokyo, Japan) and HEDS-5500#E14 two channel incre-

mental rotary encoders (Broadcom Inc., San Jose, California) are mounted to the stages. The stepper motors are two-phase with 1.8° step size. The encoders have 200 counts per revolution, giving a resolution of $2.5\text{ }\mu\text{m}$. A Kenwood PR36-3A regulated DC power supply drives the motors with 32 V. Allegro A4988 microstepping bipolar motor drivers (on a Pololu carrier board, Pololu Corporation, Las Vegas, Nevada) control and limit the current to the motor coils. The drivers are configured for half-stepping, giving a translation resolution of $5\text{ }\mu\text{m}$ per half step.

Figure 5.11 is a block diagram showing the electrical components of the system and their interconnections. The generator was modified to allow for computer-controlled x-ray exposures by connecting relays across the hand-triggered rotor drive and exposure switches, and by reading out the rotor inhibit signal. In order to synchronize x-ray exposure, flat-panel detector triggering, monitor photodiode readout, translation stage motion, and rotary encoder readout, and to perform these actions in the appropriate sequence for scatter image data acquisition, the electrical components of the system are controlled with an Arduino Due microcontroller board (Arduino, Turin, Italy). A push button on the generator switch panel was wired as an enable switch for microcontroller-initiated exposures.

The user of the system initiates scatter image scans using a custom graphical user interface (GUI) written in Python (version 3.4.4) using the PyQt package (version 5.5.1). For compatibility with the flat-panel detector's frame grabber board, the GUI runs on a PC with a 32-bit Windows XP operating system. A Python wrapper for the functions in Perkin Elmer's X-ray Imaging Software Library was written to control the settings of and read out frames from the flat-panel detector. The PC communicates with the Arduino microcontroller over a USB connection using the pySerial package (version 3.0.1) and the drivers shipped with the Arduino IDE (version 1.8.9).

The x-ray tube kVp and mA and the exposure time are set by the user on the switch panel of the generator. These values are also entered in the GUI when setting up an im-

Figure 5.11: Block diagram of the electrical components of the x-ray tube-based coherent scatter projection imaging system and their interconnections. Other than the USB connection to the PC, all I/O connections between the microcontroller and other components are via the custom-designed interface circuit.

Figure 5.12: Manufacturer-supplied anode cooling curve data for the Machlett Dynamax 69MB x-ray tube insert, with exponential fit.

age scan, and the software computes the equilibrium heat load that would be deposited in the anode if the scan continued indefinitely. The computation is based on the energy delivered per exposure, a model of the manufacturer-supplied anode cooling curve data, and the estimated time interval between exposures for detector readout, object translation, etc. Several points on the cooling curve in the manufacturer's data sheet were estimated and are plotted in Fig. 5.12. An exponential of the form $\beta_1 e^{-\beta_2 t} + \beta_3$ was fit to the data and approximates the data well when the thermal energy stored in the anode is between about 75 kJ and 220 kJ. The fit parameters are $\beta_1 = 143.8$ kJ, $\beta_2 = 0.0091 \text{ s}^{-1}$, and $\beta_3 = 73.3$ kJ.

Let $U(t)$ be the thermal energy stored in the anode as a function of time. Assuming exponential cooling, the rate of change in U is

$$U' = -\beta_2(U - \beta_3). \quad (5.1)$$

For a single-phase generator, the anode power input during exposure, in kJ s^{-1} , is given by

$$U' = 0.74 \text{ kVp} \times \text{mA}, \quad (5.2)$$

and the total rate of change of U during exposure is

$$U' = 0.74 \text{ kVp} \times \text{mA} - \beta_2(U - \beta_3). \quad (5.3)$$

The solution to this differential equation is

$$U(t) = \frac{0.74 \text{ kVp} \times \text{mA}}{\beta_2} (1 - e^{-\beta_2 t}) + \beta_3 (1 - e^{-\beta_2 t}) + U_0 e^{-\beta_2 t}, \quad (5.4)$$

where $U_0 \equiv U(t = 0)$. This solution is physically sensible for $U_0 > \beta_3$.

During a scatter image scan there are repeated cycles of a period of no exposure of duration Δt_C (while the detector is being read out, the translation stages are moving, etc.) followed by a period of exposure of duration Δt_E . If the scan is started when there is relatively little thermal energy stored in the anode then these repeated cycles of cooling and exposure will cause the stored thermal energy to first increase and then level off at some equilibrium value.

Let U_{0C} and U_{0E} be the thermal energy stored in the anode at the start of the cooling and exposure portions of a cycle, respectively. Then,

$$U_{0E} = U(\Delta t_C) = \beta_3 (1 - e^{-\beta_2 \Delta t_C}) + U_{0C} e^{-\beta_2 \Delta t_C}, \quad (5.5)$$

and the thermal energy stored in the anode at the end of a cycle is given by

$$U(\Delta t_C + \Delta t_E) = \frac{0.74 \text{ kVp} \times \text{mA}}{\beta_2} (1 - e^{-\beta_2 \Delta t_E}) + \beta_3 (1 - e^{-\beta_2 \Delta t_E}) + [\beta_3 (1 - e^{-\beta_2 \Delta t_C}) + U_{0C} e^{-\beta_2 \Delta t_C}] e^{-\beta_2 \Delta t_E}. \quad (5.6)$$

At equilibrium, the thermal energy stored in the anode at the end of a cycle, $U(\Delta t_C + \Delta t_E)$, is equal to that at the start of the cycle, U_{0C} . Substituting U_{eq} for $U(\Delta t_C + \Delta t_E)$ and U_{0C} in Eq. (5.6) yields

$$U_{eq} = \beta_3 + \frac{0.74 \text{ kVp} \times \text{mA}}{\beta_2} \left(\frac{1 - e^{-\beta_2 \Delta t_E}}{1 - e^{-\beta_2 (\Delta t_E + \Delta t_C)}} \right). \quad (5.7)$$

The user can specify an additional delay time between exposures to increase Δt_C and decrease U_{eq} . If the computed equilibrium heat load exceeds 185 kJ (5/6 of anode capacity), the scan will not proceed. Requiring the equilibrium heat load to be below an upper limit is a conservative restriction for image scans because the instantaneous thermal energy stored in the anode may never reach the equilibrium level during an image scan. The equilibrium heat load is, however, simple to compute since the past history of the tube does not need to be taken into account.

5.3 Data acquisition and processing

In addition to entering the kVp, mA, and exposure time in the GUI, the user also enters the travel speeds of the vertical and horizontal translation stages, the number of rows and columns of pixels to acquire per pencil beam (the total number of pixels in the final image is the product of the rows, columns, and number of beams), the spacings between the rows and between the columns, and any additional delay time between exposures. When using

multiple beams, the number of rows and columns and their spacings should be such that there are no overlaps or gaps between the regions of the object sampled by the beams. The trajectory of the object during the scan was programmed such that pixel data are acquired row by row from top to bottom, odd rows are acquired left to right viewed from the source, and even rows right to left. The pixels are acquired in a rectangular array. In principle, the stage motions can be programmed to execute any trajectory and to acquire pixels in any array shape.

When the scan is started, the image scan parameters are saved to a text file (JSON format) and then a sequence of alternating dark and exposure cycles is executed until the data for all of the rows and columns are acquired. Both the dark and exposure cycles follow the same sequence of events, with exceptions described in the following paragraph: (1) the stages advance to the next position, (2) the rotary encoder positions are read and appended to a text file, (3) the microcontroller waits for half of the specified additional delay time between exposures, (4) the rotor drive switch is closed, (5) the microcontroller waits until the rotor inhibit signal is cleared, (6) the detector is triggered to read out the photodiode array without transferring a frame to the PC (in order to clear out any signal that has accumulated since the last readout), (7) the exposure switch is closed for the duration of the exposure time and during exposure the output of the monitor diode charge amplifier is sampled every 200 μs by an analog input of the microcontroller, and the three largest samples are stored, (8) the exposure switch is opened, (9) the monitor diode charge amplifier is reset, (10) the detector is triggered to be read out with frame transfer to a buffer on the PC, and during this time the rotor drive switch is temporarily opened, (11) the detector frame is saved to an uncompressed 16-bit greyscale TIFF file, (12) the stored monitor diode samples are averaged and appended to a text file. The user can pause or terminate the scan at any time. After the last exposure cycle the stages return to their starting positions.

The difference between the dark and exposure cycles is that for dark cycles, the av-

eraged monitor diode samples are not appended to a text file and the exposure switch is not closed, so that there is no x-ray exposure. For exposure cycles, all the stepping signals are transmitted from the microcontroller to the motor driver boards but the transfer of power from the motor power supply to the stepper motors is disabled, so that the translation stages do not move. Furthermore, the rotary encoder readings are not appended to a text file during an exposure cycle.

The reason why all the same actions are performed during the dark cycle as during the exposure cycle, rather than just detector readout, is that the dark signal of the detector is sensitive to the timing pattern of detector frame readouts, and so the dark frame must be read out in exactly the same pattern as the exposure frame. This sensitivity also causes a rapid change in the dark signal when the image scan begins, so a set of ten dummy dark and exposure cycles is first made at the start of an image scan to allow the dark signal to settle into the timing pattern of the scan. For the dummy cycles, all stage motion and x-ray tube exposure is disabled, and no text files are appended to.

The scan data consist of a dark frame, an exposure frame, a monitor diode reading, and rotary encoder readings for each stage position. The dark frame acquired prior to each exposure frame is used as the dark signal to be subtracted from that frame. The reason this is done rather than, for example, taking the last dummy frame as the dark signal to be subtracted from all exposure frames (and thereby cutting the overall acquisition time in half) is because there is a slow and unpredictable drift in the detector dark signal, which, if not tracked over time, would result in a noticeable drift in the scatter pixel values for scans comprising hundreds of exposures.

For the x-ray tube-based scatter imaging system, correction of a scatter pattern consists only of dark frame subtraction, replacing the values of defective detector elements by the means of their neighbours, and normalizing by the associated monitor diode reading. Unlike for the synchrotron-based system, for the x-ray tube-based system a room frame is not

acquired because the x-ray intensity beyond the pencil beams measured by the flat-panel detector when no object is in the beams was found to be negligible. In the lab there are no motorized translation stages for the flat-panel detector so the scanning method (Sec. 3.3.1) cannot be employed for a true flat-field correction. It is assumed that, due to the anode heel effect, the spread in the exposure variation factor of the x-ray tube beam is greater than the spread in the true sensitivities of the detector elements, so performing no flat-field correction is preferred to performing the usual correction method.

For each exposure, the corrected frame consists of a set of overlapping Laue rings. The overlapping rings are disentangled into a set of scatter profiles (one profile per beam) using an iterative MLEM algorithm (Sec. 3.3.3). The scatter profiles are discretized into bins 0.2° wide, the centre of the first bin being 0.1° , and the number of bins is determined by the maximum angle of scatter out of a pencil beam to a detector element. To use the MLEM algorithm, a system matrix mapping the elements of the scatter profiles to the elements of the flat-panel detector needs to be constructed. For pencil beams not parallel to the system axis, a cone of scatter having a fixed angle will yield an ellipse when it intersects the detector.

The construction of the system matrix is based on the geometry of Fig. 5.13. All of the scatter from the pencil beam under consideration is assumed to originate from point P in the figure, where the centreline of the beam intersects the midplane of the object. The centre of the pencil beam is known a priori to intersect the detector plane at a point having coordinates (x_0, y_0) with respect to where the system axis intersects the detector. For an arbitrary detector element having coordinates (x, y) , it is desired to determine the angle, θ , through which a photon scatters out of the beam in order to reach that element. Taking the dot product of the pencil beam direction and the scattered photon trajectory, the angle can be found via

Figure 5.13: Geometry for x-ray tube-based system matrix construction.

$$\cos \theta = \frac{x_0 \left(x - x_0 \frac{s}{s+d}\right) + y_0 \left(y - y_0 \frac{s}{s+d}\right) + (s+d)d}{\sqrt{x_0^2 + y_0^2 + (s+d)^2} \sqrt{\left(x - x_0 \frac{s}{s+d}\right)^2 + \left(y - y_0 \frac{s}{s+d}\right)^2 + d^2}}. \quad (5.8)$$

The scattering angle to this detector element is calculated for each beam and it falls uniquely into one angular bin of the beam's scatter profile. The detector element is thus linked to exactly one element of each scatter profile. The system matrix elements joining these scatter profile elements to the detector element are assigned a value of 1, and the system matrix elements joining all other scatter profile elements to the detector element are assigned a value of 0. The MLEM algorithm attempts to divide the intensity measured by the detector element among its linked scatter profile elements and to do so in a self-consistent manner wherein the contributions of a profile element to all of its linked detector elements are the same. The value of a profile element is effectively computed as an average of its estimated contributions to its linked detector elements. The system matrix consists of mostly 0's and a few 1's, and sparse matrix methods are used to implement the MLEM algorithm. Detector elements behind primary attenuators are excluded from the procedure.

After the scatter profiles are computed for each beam location on the object, integrating

them over a range of θ gives a single value for each pixel in the scatter image. For the primary image, the pixel value for a particular pencil beam is computed as the mean of the 80 most intense element values behind the attenuator disc in the corrected scatter pattern. It was found that when acquiring an image of a uniform object, the scatter and primary pixel values decrease when plotted against the exposure number. This is because the anode expands and shifts as it heats up over the course of many exposures and this causes the focal spot to move. The primary beam spot also moves with respect to the attenuator disc and the intensity of the beam decreases as alignment with the collimator plates breaks. Consequently, the scatter and primary x-ray intensities measured by the detector also decrease. It was also found, however, that the decreasing trend in the scatter pixel values can be corrected by normalizing by the primary pixel values.

Since the pencil beams have different intensities, an inter-beam normalization is performed. The rotary encoder readings associated with each exposure frame as well as the relative spacing of the beams are used to determine the location of each pixel. The pixel value and location information are used to construct the 2D projection image as a Voronoi diagram.

5.4 System assessment

5.4.1 Beam spectrum

An energy-integrating flat-panel detector can only measure the angle of scattered x rays and not their energy. Measurements with the x-ray tube-based system only provide scatter intensity as a function of θ and not as a function of $|\vec{x}|$, that is, the system measures scatter profiles. Consequently, using a broad spectrum beam from an x-ray tube contributes to the angular blur of the measured scatter profiles.

The spectral blur washes away the features of the scatter profiles that are characteristic

of different materials, reducing contrast in the scatter image. Therefore, contrast improvement is expected if a narrower spectrum of x rays is used to irradiate the object. A narrower spectrum can be achieved by using a lower x-ray tube voltage and by using filters such as aluminum, which attenuates low energy photons, K-edge filters, which attenuate photons with energy just above the K-shell binding energy of the filter atoms, or balanced filters, which give a pseudomonoenergetic spectrum after subtracting two acquisitions. Using a lower tube voltage and filters, however, reduces the number of photons incident on the object and subsequently the number of scattered photons incident on the detector, which increases the relative importance of the noise in the pixel values.

The distinguishability of two materials in the image can be quantified by the signal-difference-to-noise ratio (SDNR), which is computed as

$$\text{SDNR} = \frac{|\bar{p}_1 - \bar{p}_2|}{\sqrt{s_1^2 + s_2^2}}, \quad (5.9)$$

where $\bar{p}_1$ and $\bar{p}_2$ are the means of pixel values in regions of interest in the first and second materials, respectively, and s_1 and s_2 are the sample standard deviations of those pixel values. To study the compromise between scatter profile sharpness and intensity, the SDNR was computed for several scatter images of an object consisting of 1.0 cm thick polyethylene and polystyrene blocks. The object midplane was positioned 35.0 cm from the focal spot and each image was acquired with different combinations of tube voltage and beam filters.

The images were single-beam acquisitions using only the middle pencil beam of the fifteen-beam array. The tube voltage and exposure time were adjusted in tandem to maintain the same anode heat loading per exposure for each image acquisition. The tube current for each acquisition was 200 mA. For each tube voltage and filter combination, the relative air kerma of the uncollimated x-ray beam was measured using an ionization chamber

Table 5.2: Materials, thicknesses, and K-shell binding energies²⁰⁶ of the filters used for beam spectra studies.

Material	Thickness (mm)	K-shell binding energy (keV)
Al	2.25	1.6
Sn	0.025	29.2
Ho	0.127	55.6
W	0.100	69.5

(PRM model D-15, Precision Radiation Measurement Inc., Nashville, Tennessee) placed at the object position. The different filter materials used, their thicknesses, and their K-shell binding energies are listed in Table 5.2.

5.4.2 Air kerma

The air kerma at 35.0 cm from the source was measured for each of the fifteen pencil beams using radiochromic film (Gafchromic XR-QA2, Ashland, Inc., Bridgewater, New Jersey) for a 33.3 mAs exposure with a 110 kVp beam filtered with 2.25 mm Al. Two air kerma measurements were made, one with all fifteen beams simultaneously irradiating the film and one with only the middle beam of the fifteen-beam array. The film was affixed to a block of polystyrene foam and placed on the translation stage platform so that it was standing perpendicular to the system axis and facing the source. The irradiated films were scanned to 48-bit colour TIFF files at 300 dpi with no colour corrections or filters using a Canon CanoScan LiDE 60 scanner (Canon Inc., Tokyo, Japan).

A 19-point scanned-pixel-red-channel-value to air kerma calibration set spanning 0–100 mGy was acquired for the film batch and scanner. The calibration films were exposed at 110 kV using an X-Rad 320 cabinet x-ray system (Precision X-Ray, North Branford, Connecticut) and the air kerma for the exposures was measured using a 0.6 cm³ PTW 30010 ionization chamber with a PTW UNIDOS T10002 electrometer (PTW-Freiburg,

Freiburg, Germany). The calibration procedure of Rampado et al.^{207,208} was performed using a second-order polynomial fit to the calibration data.

For most image scans, the spacing between pixels is less than the diameter of the pencil beams and so the cumulative air kerma incident on a point of the object during an image acquisition is higher than the air kerma delivered at the centre of a single beam during one exposure. To measure the average total air kerma incident on an object, image scans of approximately 1 cm × 1 cm regions of radiochromic film were made with four different pixel spacings, using only the middle pencil beam at 110 kVp with 2.25 mm Al.

5.4.3 Signal-to-noise ratio

Six single-beam scatter images of an object made up of 1.0 cm thick polyethylene and polystyrene blocks were acquired with the object midplane positioned 35.0 cm from the source. The images were acquired at different exposure settings ranging from 1.7–10.0 mAs and using a 110 kVp beam filtered with 2.25 mm Al. The beam air kerma for each acquisition was determined by scaling the middle beam's air kerma measurement for a 33.3 mAs exposure with the same beam spectrum (Sec. 6.2) to the mAs of the acquisitions.

The polyethylene-polystyrene SDNR was computed according to Eq. 5.9 for each image using 32-pixel regions within each block. The pixel SNR for a 56-pixel region within the polystyrene block was also computed by dividing the mean of the pixel values by their sample standard deviation. The SDNR and SNR were computed for the primary images and the scatter images, with scatter profiles integrated $2.0^\circ \leq \theta \leq 3.0^\circ$.

5.4.4 Edge sharpness

As for the synchrotron-based system (Sec. 3.4.2), the spatial resolution of the x-ray tube-based system was determined by measuring the width of sharp edges in the scatter images.

Acquisitions were made of an object made up of 1.2 cm thick polyethylene and polycarbonate blocks, one placed above the other. The midplane of the object was positioned 35.0 cm from the source and images were acquired with different pixel spacings. All images were acquired with 6.7 mAs exposure per scatter pattern and with a 110 kVp beam filtered with 2.25 mm Al. Single-beam and fifteen-beam images were acquired. For the fifteen-beam acquisitions, various plastic blocks 1.0–1.5 cm thick were built up around the central polyethylene and polycarbonate blocks to make an 8 cm $\times$ 11 cm object. The vertical width of the horizontal boundary between the polyethylene and polycarbonate blocks was determined for each image. The middle beam of the fifteen-beam array probed the horizontal boundary of interest for all acquisitions.

Similar to Eq. (3.18), an ideal horizontal boundary in the image at location $y = y_0$ between two materials with pixel value A_B below the boundary and pixel value A_A above the boundary can be modelled by

$$E_0(x, y) = (A_A - A_B)H(y - y_0) + A_B, \quad (5.10)$$

where again, $H(y)$ is the Heaviside unit-step function and $E_0(x, y)$ is constant along x .

From the scan of the radiochromic film irradiated by all fifteen beams (Sec. 5.4.2), each beam has a disc of roughly constant intensity at the centre and a penumbra around the disc decreasing to zero intensity. The penumbra region can be assumed to have linear decrease in intensity, and so the pencil beam intensity distribution is approximately a conical frustum with smaller radius r_1 and larger radius r_2 . The impulse response function for aperture blur of a boundary, normalized to unit volume, can be modelled as

$$h(x, y) = \frac{3}{\pi(r_1^2 + r_2^2 + r_1 r_2)} \frac{1}{(r_2 - r_1)} \times \left[(r_2 - r) \text{Rect}\left(\frac{r}{2r_2}\right) - (r_1 - r) \text{Rect}\left(\frac{r}{2r_1}\right) \right], \quad (5.11)$$

where $r = \sqrt{x^2 + y^2}$.

A real horizontal boundary is given by

$$\begin{aligned}
E(x, y) &= E_0(x, y) ** h(x, y) \\
&= \frac{3(A_A - A_B)}{\pi(r_1^2 + r_2^2 + r_1 r_2)} \frac{1}{(r_2 - r_1)} \\
&\quad \times \operatorname{Re} \left\{ \frac{1}{3} (y - y_0)^3 \left(\cosh^{-1} \left(\frac{r_1}{y - y_0} \right) - \cosh^{-1} \left(\frac{r_2}{y - y_0} \right) \right) \right. \\
&\quad - \frac{2}{3} (y - y_0) \left(r_1 \sqrt{r_1^2 - (y - y_0)^2} - r_2 \sqrt{r_2^2 - (y - y_0)^2} \right) \\
&\quad + \frac{1}{6} \left(r_1^3 \tan^{-1} \left(\frac{y - y_0}{\sqrt{r_1^2 - (y - y_0)^2}} \right) - r_2^3 \tan^{-1} \left(\frac{y - y_0}{\sqrt{r_2^2 - (y - y_0)^2}} \right) \right) \\
&\quad \left. - \frac{1}{2} \left(r_1^3 \sin^{-1} \left(\frac{y - y_0}{r_1} \right) - r_2^3 \sin^{-1} \left(\frac{y - y_0}{r_2} \right) \right) \right\} + \frac{A_A + A_B}{2}, \quad (5.12)
\end{aligned}$$

where $**$ denotes two-dimensional convolution. $E(x, y)$ is constant in x . In the y direction, $E(x, y)$ is equal to the constant A_B for $(y - y_0) \leq -r_2$ and is equal to the constant A_A for $(y - y_0) \geq r_2$. For $-r_2 < (y - y_0) < r_2$, the function transitions smoothly from A_B to A_A , so the blurred horizontal boundary can be said to have vertical width $W_y = 2r_2$.

When determining the vertical width of the horizontal boundary between the polyethylene and polycarbonate blocks, Eq. (5.12) was fit to the values of the pixels ordered from below to above the boundary for each column of the image containing the two blocks. To compute the boundary width for an image, the values for the fit parameter r_2 from each column fit were averaged, and the uncertainty was computed as the standard deviation of the mean of the values.

5.4.5 Object thickness

The capability of the system to image objects of different thickness was also studied. Starting from a very thin object, increasing the thickness increases the number of coherent scatter interactions and therefore the intensity of scattered x rays at the detector, but the attenuation of both primary and scattered photons also increases and eventually overtakes scatter photon generation. In theory, for an object of uniform composition, the maximum scatter intensity downstream of the irradiated object is for a thickness of $1/\mu$, where μ is the linear attenuation coefficient of the object's material. Luggar et al.,²⁸ however, determined that in practice the maximum scatter intensity is for object thickness $\approx \frac{2}{3\mu}$.

With no collimation between the object and detector, however, the angular blur of the scatter patterns progressively increases with thickness since the scatter originating at different depths in the object superposes at the detector plane. This thickness blurring of scatter profiles reduces material contrast and so the object thickness that maximizes SDNR may be somewhat less than that which maximizes scatter intensity, depending on the relative importance of the thickness blur.

Two experiments were performed to assess the effect of object thickness on scatter image SDNR. The first experiment investigated the distinguishability of target materials embedded in a thick background. The object was made up of a 1.0 cm thick block of polystyrene placed on top of a 1.0 cm thick block of polyethylene and its midplane was positioned 35.0 cm from the source. Several single-beam image acquisitions were made with varying thickness of PMMA blocks symmetrically added in front of and behind the polyethylene and polystyrene blocks, for a total PMMA thickness ranging from 0–8.4 cm. The second experiment investigated the impact of the thickness of the target materials themselves, free of background. The thickness of the polyethylene and polystyrene blocks was varied from 1.0–5.0 cm and no additional material was added.

All images were acquired with 6.7 mAs exposure per scatter pattern and with a 110 kVp

beam filtered with 2.25 mm Al. The polyethylene-polystyrene SDNR was computed according to Eq. (5.9) for each scatter image using 32-pixel regions within each block. For each acquisition, scatter profiles were integrated over an angular range that maximized SDNR for that image.

5.4.6 Object-to-detector distance

For a fixed source-to-detector distance of 47.5 cm, moving the object closer to the detector increases the measured scatter intensity because the fixed-size detector elements intercept photons from a larger range of scattering angles, but this reduces the angular resolution of the scatter profiles. This trade-off between profile resolution and intensity suggests that there is an optimal distance from the detector at which the object should be placed.

A 1.0 cm thick object made up of polyethylene and polystyrene blocks was placed at different object-midplane-to-detector distances and single-beam scatter images were acquired with 6.7 mAs exposures using a 110 kVp beam filtered with 2.25 mm Al. The polyethylene-polystyrene SDNR was computed according to Eq. (5.9) for each scatter image using 32-pixel regions within each block. For each acquisition, scatter profiles were integrated over an angular range that maximized SDNR for that image.

5.4.7 Contrast-detail evaluation

For the contrast-detail evaluation, the same phantom that was imaged using the synchrotron-based system (Sec. 3.4.3) was used. Single-beam and fifteen-beam scatter images were acquired with 6.7 mAs exposure per scatter pattern and with a 110 kVp beam filtered with 2.25 mm Al, for which the average energy of the beam is approximately 50 keV and $\mu \approx 0.25 \text{ cm}^{-1}$ for PMMA.¹⁹⁵ The pixel spacing was 0.8 mm for each image.

As for the synchrotron-based system, for each image of the phantom, for every disc di-

ameter an observer decided which thickness PMMA disc was just distinguishable from the background and the contrast of that disc was determined using Eq. (3.28). To compare the contrast-detail images acquired with the x-ray tube-based system to those acquired with the synchrotron-based system, the minimum detected contrasts were normalized to a reference object-incident air kerma, $K_{air,ref}$, by scaling by $(K_{air}/K_{air,ref})^k$. The value of k was determined for the x-ray tube-based system in Sec. 6.3 and for the synchrotron-based system in Sec. 4.5.1. These normalized contrasts for both the x-ray tube-based system acquisitions and the synchrotron-based system acquisitions were plotted against disc diameter to generate contrast-detail curves.

Chapter 6

X-ray tube-based system: Images and results

This chapter presents results related to the x-ray tube-based coherent scatter projection imaging system. They include comparison of beam spectra, imaging air kerma measurements, scatter images, and experimentally-determined image quality metrics.

6.1 Choice of beam spectrum

Several scatter image acquisitions of PE and PS blocks were made to determine a suitable x-ray beam spectrum. For each scatter image, Table 6.1 lists the tube voltage and filter combination, the exposure time per scatter pattern, the relative air kerma of the beam at the object, and the polyethylene-polystyrene SDNR computed from the image data. Regions of 32 pixels in each of the polyethylene and polystyrene blocks were used to compute the SDNR. The angular range over which scatter profiles were integrated to generate pixel values was chosen specific to each image to give the maximum SDNR.

Also included in Table 6.1 are the SDNR normalized by the relative K_{air} and the SDNR

Table 6.1: SDNR computed for single-beam scatter images of polyethylene and polystyrene blocks acquired using different tube voltage and filter combinations. Each acquisition had tube current 200 mA.

Tube voltage (kV)	Filter	Exposure time (ms)	Relative K_{air}	SDNR	SDNR/ K_{air}	SDNR/ $\sqrt{K_{air}}$
110	none	33.3	1.00	26.8 ± 2.4	26.8 ± 2.4	26.8 ± 2.4
88	none	41.6	0.89	21.0 ± 2.0	23.6 ± 2.2	22.3 ± 2.1
73	none	50.0	0.81	12.8 ± 1.4	15.8 ± 1.7	14.2 ± 1.5
110	Al	33.3	0.52	19.9 ± 1.8	38 ± 3	27.5 ± 2.5
88	Al	41.6	0.40	17.3 ± 1.7	43 ± 4	27.2 ± 2.7
73	Al	50.0	0.33	12.1 ± 1.2	37 ± 4	21.2 ± 2.0
110	Sn	33.3	0.76	27.4 ± 2.8	36 ± 4	32 ± 3
88	Sn	41.6	0.65	20.0 ± 2.1	31 ± 3	24.8 ± 2.6
73	Sn	50.0	0.58	10.8 ± 1.2	18.6 ± 2.0	14.2 ± 1.6
110	Ho	33.3	0.23	14.2 ± 1.3	60 ± 6	29.3 ± 2.7
88	Ho	41.6	0.17	14.5 ± 1.6	84 ± 9	35 ± 4
73	Ho	50.0	0.13	6.1 ± 0.7	46 ± 5	16.8 ± 1.9
110	W	33.3	0.13	11.3 ± 1.1	88 ± 9	31 ± 3
88	W	41.6	0.08	9.3 ± 0.8	119 ± 11	33 ± 3
73	W	50.0	0.05	4.8 ± 0.5	102 ± 11	22.1 ± 2.4

normalized by the square root of the relative K_{air} . The unnormalized SDNR is a practical measure that compares the performance of the system for the same tube loading. SDNR/ K_{air} is a measure of the SDNR performance for the same object dose. The SDNR is expected to increase as $\sqrt{K_{air}}$ and so SDNR/ $\sqrt{K_{air}}$ is a measure of how efficiently the available photons are used. There is no decidedly best tube voltage and filter combination but 110 kV with 2.25 mm Al was chosen as the default spectrum to be used for image acquisitions since it performs well for all three measures. A 110 kV beam filtered with 0.025 mm Sn would also be suitable but the thicker Al plate is easier to handle than the very thin Sn foil. Use of lower tube voltages would result in insufficient primary transmission through the 1.5 mm thick attenuator discs. The W-filtered beams give very high SDNR per unit dose to the object but the low x-ray fluence post-filter means that about four times the tube loading is needed to achieve the same SDNR as with the Al-filtered beams.

Table 6.2: Average air kerma in mGy at the object midplane within each pencil beam of the x-ray tube-based system for a 33.3 mAs exposure with a 110 kVp beam filtered with 2.25 mm Al, measured using radiochromic film. Column 1 is the left-most column of beams, viewed from the source.

Beam row \ Beam col.	1	2	3	4	5
1	6.8 ± 1.0	7.1 ± 1.1	7.0 ± 1.1	7.3 ± 1.1	6.9 ± 1.1
2	8.3 ± 1.2	9.3 ± 1.2	9.3 ± 1.2	9.0 ± 1.2	8.5 ± 1.2
3	9.4 ± 1.2	9.9 ± 1.3	9.6 ± 1.2	8.9 ± 1.2	7.8 ± 1.1

6.2 Air kerma

For the film exposed using only the middle pencil beam of the fifteen-beam array, the average air kerma within the beam for a 33.3 mAs exposure at 110 kVp filtered with 2.25 mm Al was measured to be 9.7 ± 1.3 mGy. For the film exposed using all fifteen beams, the measured average air kerma for each beam is given in Table 6.2. The top row of beams, those which make the largest angle relative to the sloping face of the x-ray tube anode, have the lowest average air kerma.

Image scans of the film made with 0.5 mm and 0.8 mm pixel spacing were acquired with 6.7 mAs per exposure and scans made with 1.0 mm and 1.5 mm pixel spacing were acquired with 13.3 mAs per exposure. For larger pixel spacings, the partial beam overlap gives texture in the air kerma distribution at the object (Fig. 6.1). For smaller pixel spacings the air kerma distribution is more uniform. The average total air kerma within the $1 \text{ cm} \times 1 \text{ cm}$ exposed regions was computed and scaled up to 33.3 mAs exposures. The scaled average air kerma was divided by the average air kerma of the middle pencil beam and is plotted against pixel spacing in Fig. 6.2.

There is an inverse relationship between average total incident air kerma and pixel spacing. The average object air kerma approaches infinity as the pixel spacing goes to zero. For 6.7 mAs exposure per scatter pattern using a 110 kVp beam filtered with 2.25 mm Al

Figure 6.1: Object air kerma distribution (mGy) of a single-beam scatter image acquisition with 1.5 mm pixel spacing, measured using radiochromic film. Seven rows and seven columns of exposures were made. The beam diameter is ≈ 3.4 mm and the areas of the film irradiated by three adjacent exposures are indicated by the white circles. The film was scanned at 300 dpi.

and for a pixel spacing of 0.5 mm, the average object-incident air kerma was measured to be 45 ± 3 mGy. Reducing the pixel spacing increases the imaging time and dose to the object but with little gain in spatial resolution, which becomes limited by the beam size rather than the spatial sampling interval (Sec. 6.5). There is also no gain in SNR since the number of photons incident on the detector per scatter pattern is independent of pixel spacing.

6.3 Signal-to-noise ratio

Figure 6.3 is a log-log plot of the polyethylene-polystyrene SDNR computed for the single-beam scatter and primary images acquired for different object-incident air kerma. Likewise,

Figure 6.2: Average air kerma incident on a $1\text{ cm} \times 1\text{ cm}$ region of radiochromic film, relative to the air kerma delivered within the middle pencil beam during one exposure, for a single-beam scatter image acquisition with different pixel spacings.

Fig. 6.4 is a plot of the polystyrene pixel SNR. Power laws were fit to each data set giving $SDNR \propto K_{air}^{1.4 \pm 0.6}$ and $SNR \propto K_{air}^{0.99 \pm 0.11}$ for primary, and $SDNR \propto K_{air}^{0.68 \pm 0.09}$ and $SNR \propto K_{air}^{0.63 \pm 0.09}$ for scatter.

For the primary images of the plastic blocks, the exponents of the power-law fits indicate that the image is readout-noise limited. This requires further investigation. Low transmission of the primary beam through the W / Cu attenuator discs would cause the readout noise to dominate over x-ray photon statistical noise. The primary signal, however, is well above the flat panel's lower detection limit. The SDNR is very low even for 3 mGy exposures, largely due to the poor polyethylene-polystyrene contrast for primary x-ray imaging. In general, the primary images acquired using the x-ray tube-based system were found to be of very poor quality. These results suggest that rather than using the same detector to measure both the scattered and primary photons by attenuating the transmitted

Figure 6.3: Polyethylene-polystyrene SDNR computed for the single-beam scatter and primary images acquired for different object-incident air kerma.

Figure 6.4: Polystyrene pixel SNR computed for the single-beam scatter and primary images acquired for different object-incident air kerma.

beam to meet the dynamic range of the detector, it would be better to use separate small detectors with lower sensitivity for each primary beam, as was done for the synchrotron-based system. Comparing Fig. 6.4 to Fig. 4.6, this change should improve the primary SNR by at least 1–2 orders of magnitude and make the data quantum noise limited. The main challenge would be managing the cables of the fifteen small detectors so that they do not disturb the x rays scattered by the object or generate their own scatter.

The exponents for the power-law fits to the scatter image SDNR and SNR data indicate that the scatter data are closer to being quantum-noise limited and that the scattered photon intensity incident on the flat-panel detector is well above the noise floor. For the sample object consisting of 1 cm thick plastic blocks, the SDNR even at the lowest beam air kerma used, 0.5 mGy (or ~ 3.6 mGy total average object-incident air kerma, taking beam overlap into account), was 8.8 ± 1.0 , so the beam intensity and dose to this particular object can be further reduced while maintaining $\text{SDNR} > 5$. Having a signal difference five times larger than the noise ensures that there is a vanishing probability of confusing one material's signal with another material's due to noise fluctuations.^{209,210} This is known as the Rose criterion. Comparing Fig. 6.4 to Fig. 4.6, both the x-ray tube- and synchrotron-based systems have scatter image SNR in the range of 10–100, however, the x-ray tube system measurements were for lower object-incident air kerma, indicating improved scatter SNR performance, likely due to the higher sensitivity of the flat-panel detector.

6.4 Images

Figure 6.5 (a) is a fifteen-beam scatter image acquisition of a 1.0 cm thick object consisting of blocks of polyethylene, polystyrene, polycarbonate, PMMA, and nylon. The object midplane was placed 35.0 cm from the source. Scatter profiles were integrated $2.0^\circ \leq \theta \leq 3.0^\circ$. The image was acquired with 1.7 mAs exposure per scatter pattern and with a

Figure 6.5: (a) Fifteen-beam scatter image and (b) conventional primary image of a 1.0 cm thick object consisting of plastic blocks. The primary image was demagnified from the detector plane to the object plane. The labelled blocks are polyethylene (PE) and polystyrene (PS).

110 kVp beam filtered with 2.25 mm Al. A $6.0\text{ cm} \times 10.0\text{ cm}$ region ($2.0\text{ cm} \times 2.0\text{ cm}$ per beam) was acquired with 1.0 mm pixel spacing. The average total air kerma incident on the object during the acquisition was 3.1 mGy. Acquisitions were made with no additional delay between exposures and the image acquisition time was 8.8 min for 6000 pixels. This is currently the shortest acquisition time achievable with the system for an image of this size and pixel spacing. If the flat-panel detector's dark signal drift was negligible over the course of the image acquisition then the acquisition of separate dark frames for each exposure could be omitted and the image acquisition time would be halved.

Figure 6.5 (b) is a conventional primary image of the same plastics object, acquired with all collimators and attenuator discs removed and without an anti-scatter grid. The image was acquired using a 0.3 mAs exposure and an 86 kVp beam with 2.25 mm Al. While the primary image has much better spatial resolution, being limited mainly by the element pitch of the flat-panel detector ($400\text{ }\mu\text{m}$), and an acquisition time of less than 1 s, the scatter image exhibits higher contrast. For example, the contrast of the labelled polyethylene and polystyrene blocks in Fig 6.5 (b), computed using Eq. (4.1), is 0.0098 ± 0.0025 for the primary image and 0.95 ± 0.17 for the scatter image.

Figure 6.6 (a) is a photograph of the source-facing side of a 2.5 cm thick cut of bovine caudal vertebra with surrounding tissue, commonly known as beef oxtail, that was imaged with the system. The oxtail was placed in a slab-sided polyvinyl chloride container filled with PBS. The walls of the container were 0.8 mm thick, and it had a height of 7.0 cm and a width of 9.9 cm.

Figure 6.6 (b) is a fifteen beam scatter image acquisition of the piece of oxtail, whose midplane was positioned 35.0 cm from the source. Since the sampled image width is 10 cm for fifteen-beam acquisitions for an object 35.0 cm from the source, blocks of plastic 1.0 cm thick were placed beside the container to ensure that air was not imaged. A $6.0\text{ cm} \times 10.0\text{ cm}$ region was acquired with 1.0 mm pixel spacing. The image was ac-

Figure 6.6: (a) Photograph of a cut of beef oxtail placed in a plastic container filled with PBS. (b) Fifteen-beam scatter image of the oxtail. The sampling grid with 1.0 mm pixel spacing was linearly interpolated to a finer grid with 0.1 mm spacing to smooth the pixelation. Regions of fat and muscle are indicated. (c) Conventional primary image, demagnified from the detector plane to the object plane.

quired with 6.7 mAs exposure per scatter pattern and with a 110 kVp beam filtered with 2.25 mm Al. The average total air kerma incident on the object during the acquisition was 12.4 mGy. An additional delay of 700 ms was added between exposures and the image acquisition time was 13.7 min. Scatter profiles were integrated $2.0^\circ \leq \theta \leq 3.0^\circ$.

Figure 6.6 (c) is a conventional primary image of the piece of oxtail, acquired with all collimators and attenuator discs removed and without an anti-scatter grid. The image was acquired using a 0.3 mAs exposure and an 86 kVp beam with 2.25 mm Al. Bony structures are very apparent in the primary image but there is not much contrast between fat, muscle, and PBS. In the scatter image for the angular range used, fatty regions, including the central marrow, are very bright and are easily distinguished from muscle. The fat-muscle contrast, computed using Eq. (4.1) for $3.0 \text{ mm} \times 3.0 \text{ mm}$ regions of fat and muscle within the areas indicated in Fig. 6.6 (b), is 0.100 ± 0.004 for the primary image and 0.89 ± 0.07 for the scatter image. Surrounding the central bright marrow region is a dark ring, possibly indicating the presence of cortical bone as opposed to the trabecular bone in the other bony regions.

6.5 Edge sharpness

For measurement of edge sharpness, five single-beam and five fifteen-beam scatter image acquisitions were made for the object made up of polyethylene and polycarbonate blocks (Sec. 5.4.4). The acquisition parameters as well as the computed vertical width of the horizontal boundary between the two blocks are listed in Table 6.3. Scatter profiles were integrated $2.0^\circ \leq \theta \leq 3.0^\circ$. Figure 6.7 shows one example of the transition in pixel values across the horizontal boundary along with a fit of Eq. (5.12) for the single-beam, 0.5 mm pixel spacing image.

The mean vertical width for all of the single-beam images in Table 6.3 is

Table 6.3: Vertical widths for x-ray tube-based scatter images with different acquisition parameters. Images 2 and 7 were acquired with the vertical starting position of the object offset by 0.5 mm relative to the starting positions for images 1 and 6, respectively. The number of columns per beam determines the number of boundary fits per image. The number of rows per beam determines the number of data points for each boundary fit.

Image #	Beams	Pixel spacing (mm)	Rows per beam	Cols. per beam	W_y (mm)
1	1	1.00	11	11	3.36 ± 0.14
2	1	1.00	11	11	3.20 ± 0.14
3	1	0.50	21	9	3.18 ± 0.11
4	1	0.25	41	9	3.10 ± 0.13
5	1	0.10	81	5	3.06 ± 0.13
6	15	1.00	11	11	3.56 ± 0.13
7	15	1.00	11	11	3.57 ± 0.20
8	15	0.50	21	9	3.18 ± 0.15
9	15	0.25	41	9	3.32 ± 0.13
10	15	0.10	81	5	3.54 ± 0.15

Figure 6.7: Pixel value vs pixel centre vertical position for one column in Image 3 of Table 6.3 in the vicinity of the horizontal boundary between the polyethylene and polycarbonate blocks. Equation (5.12) was fit to the data with fit parameters shown in the figure. The vertical width for this column is $2r_2 = 3.4 \pm 0.6$ mm.

3.18 ± 0.13 mm. For the fifteen-beam images, the mean width is 3.45 ± 0.15 mm. While the difference is not statistically significant, the fifteen-beam images have somewhat wider boundaries than the single-beam images and this may indicate that MLEM disentanglement inaccuracies introduce additional spatial blur. The boundary width, W_y , is independent of the pixel spacing and, for the images in Table 6.3, is consistent with the larger diameter of the frustum of the middle pencil beam, which is about 3.4 mm, as determined from measurement with radiochromic film. Each image in Table 6.3 was acquired with pixel spacing less than half of the beam diameter. It is expected that if larger pixel spacing is used, the spatial resolution of the image will be equal to the spatial sampling interval and independent of the beam diameter.

6.6 Object thickness

Figure 6.8 is a plot of the SDNR of 1.0 cm thick polyethylene and polystyrene blocks vs thickness of added PMMA blocks. When the PMMA thickness exceeds 2 cm (total object thickness 3 cm), the SDNR drops below 5 (Rose criterion). If the objects of interest, polyethylene and polystyrene in this case, were thinner, it is expected that the total object thickness including the surrounding material beyond which the SDNR falls below 5 would be less than 3 cm.

Figure 6.9 is a plot of polyethylene-polystyrene SDNR for different thicknesses of polyethylene and polystyrene blocks. The SDNR has a maximum for a thickness of around 3 cm but is high for all thicknesses imaged and for the exposure settings used. The average energy of the spectrum used for these acquisitions, calculated using a computational model,²¹¹ is about 50 keV, for which $\mu \approx 0.2 \text{ cm}^{-1}$ for both polyethylene and polystyrene.¹⁹⁵ The thickness for which uniform-object SDNR was found to be greatest is approximately equal to the object thickness at which scatter profile intensity is expected

Figure 6.8: SDNR for 1.0 cm thick polyethylene and polystyrene blocks plotted vs total thickness of PMMA added in front and behind. The insert is the side view of the object geometry. The abscissa is $t_1 + t_2$.

Figure 6.9: Polyethylene-polystyrene SDNR as a function of the thickness of the two plastic blocks. No background material was added in front or behind the blocks.

to be at a maximum, $\frac{2}{3\mu} = 3.3$ cm. Possibly the spectral blur of scatter profiles is already so severe that additional thickness blurring has little impact on SDNR, however, conclusions are limited by the difficulty of accurately assigning a value of μ for the effective attenuation of the beam spectrum.

The range of object thickness that the x-ray tube-based scatter imaging system can image is highly dependent on the material composition and distribution, but as a general remark, the system should consistently have high SDNR for low-Z objects with thickness 1–3 cm.

6.7 Object-to-detector distance

Figure 6.10 is a plot of the SDNR vs object-midplane-to-detector distance. For the beam spectrum used, the features in the polyethylene and polystyrene scatter profiles that maximize their contrast are at small angles, $<3^\circ$, and as the object is moved closer to the detector, the SDNR drops off quickly because the smaller angles map to progressively smaller radial distances from the centre of the beam in the plane of the detector, which are blocked by the primary beam attenuator disc. As the object is moved more than 10 cm from the detector, even smaller-angle regions of the scatter profiles become accessible but the scatter intensity incident on a detector element decreases, and a maximum in the SDNR is observed for an object-to-detector distance of around 15 cm.

6.8 Contrast-detail evaluation

Figure 6.11 shows the air-kerma-normalized contrast-detail curves. Included in the plot are two curves for continuous motion images acquired with the synchrotron-based system and two curves for step-and-shoot images acquired with the x-ray tube-based system. On

Figure 6.10: Polyethylene-polystyrene SDNR vs object-midplane-to-detector distance for a fixed source-to-detector distance.

Figure 6.11: Air-kerma-normalized contrast-detail curves for synchrotron-based (continuous motion, CM) and x-ray tube-based (step-and-shoot, SS) scatter image acquisitions.

these plots, better-performing systems will have curves closer to the lower-left corner. The curves in Fig. 6.11 indicate that the contrast-detail performance of the x-ray tube-based system is about the same or somewhat better than that of the synchrotron-based system.

6.9 X-ray tube-based system summary

A coherent x-ray scatter projection imaging system employing a two-dimensional array of fifteen pencil beams was implemented using a rotating-anode x-ray tube, a high voltage generator, and an energy-integrating flat-panel detector. The system was designed for imaging samples of tissues and plastics 1–6 cm thick, 10 cm wide, 6 cm tall, and mass 1 kg. Simulations were performed to determine the parameters of the system geometry. The primary imaging capability of the system was not optimized and transmitted x rays are measured with the flat-panel detector through attenuators.

Based on measurements of SDNR in scatter images of plastics, a 110 kVp beam filtered with 2.25 mm Al was found to be a suitable spectrum for the system. For this spectrum, the average air kerma of the middle beam of the fifteen-beam array is 0.28 ± 0.04 mGy per mAs at the object. Anode heating limits x-ray tube output and images are acquired in a step-and-shoot manner. Image acquisition time is about four times that of the synchrotron system.

The width of a sharp edge in scatter images was consistent with the pencil beam diameter. The optimal SDNR when imaging plastics is for object thickness ≈ 3 cm and object-to-detector distance ≈ 15 cm. Contrast-detail performance is similar to the synchrotron-based system. SNR measurements indicated that the scatter data are close to quantum-noise limited and that the scattered photon intensity is well above the noise floor of the flat-panel detector.

Chapter 7

Conclusions and future work

7.1 Conclusions

X-ray coherent scatter imaging systems unite the techniques of x-ray radiography and x-ray diffraction, exploiting the coherent scatter cross section's dependence on the molecular structure of materials to give image contrast not possible with conventional transmission imaging. Two projection scatter imaging prototype systems were developed for this thesis with the goal of achieving short image acquisition times.

The first system used a synchrotron light source which can provide high intensity monoenergetic x rays that are ideal for prototyping. This was a continuation of earlier work in the research group and the system was optimized for imaging speed by combining continuous object motion with multibeam acquisition. The total acquisition time for a $5.0\text{ cm} \times 9.0\text{ cm}$ object with 8425 pixels was 2.25 min. Continuous object motion has an approximately thirtyfold faster acquisition time compared to step-and-shoot, and using multiple beams reduces acquisition time by a factor nearly equal to the number of beams.

Use of continuous motion increases the widths of edges in the scatter image by the product of the object translation speed and the acquisition time per pixel. This had little

effect on the images presented since the objects were made up of uniform material regions with spatial extent much larger than the blurring width, but could be important if looking for small features such as microcalcifications.

Acquiring scatter data at different beam energies between 33 and 70 keV yields nearly identical images. If only a small area detector is available or if the object is thick, then a higher energy should be used.

SNR measurements for different object-incident air kerma levels indicate that, in addition to quantum noise, the readout noise of the flat-panel detector is a limitation. A detector with lower readout noise and with greater sensitivity could improve scatter image SNR. The scatter pattern usually does not need to be acquired at high resolution and so a detector with larger elements or one capable of binning elements to reduce additive readout noise would be useful.

Using lessons learned with the synchrotron-based system, a second system was developed using a more practical rotating anode x-ray tube source. Scatter imaging with an x-ray tube comes with the additional challenges of a polyenergetic spectrum and heat loading which reduce contrast and imaging speed, respectively, compared to the synchrotron-based system. Measurement of plastic SDNR in scatter images suggested that broader spectrum beams, such as 110 kVp filtered with 2.25 mm Al, are preferred to beams such as 88 kVp filtered with 0.1 mm W which have a narrower spectrum but give much lower scattered x-ray signal. The shortest acquisition time for a $6.0\text{ cm} \times 10.0\text{ cm}$ object with 6000 pixels was 8.8 min. Acquisition speed is about four times slower with the x-ray tube-based system mainly because anode heating necessitates step-and-shoot acquisition of image data. The x-ray tube-based system does have the advantage of being able to use a two-dimensional array of pencil beams, which reduces acquisition time.

The optimal SDNR when imaging plastics is for object thickness $\approx 3\text{ cm}$ and object-to-detector distance $\approx 15\text{ cm}$. Similar results are expected when imaging soft tissues but for

bone the optimal thickness will be less than 3 cm since it is more attenuating of x rays. Contrast-detail performance of scatter images was found to be similar for the x-ray tube- and synchrotron-based systems, regardless of acquisition mode. The flat-panel detector used with the x-ray tube-based system is about 100 times more sensitive than the detector used with the synchrotron-based system, and the electronic noise was much less important than the quantum noise of the scattered x rays. The difference in sensitivity between the two detectors is likely due to different design parameters for the amplifier and ADC electronics, chosen as appropriate for the intended application.

Returning to the hypothesis stated in Sec. 1.3, it is possible to implement a coherent scatter projection imaging system to image a roughly $10\text{ cm} \times 10\text{ cm}$ area in less than ten minutes and provide soft-material contrast superior to conventional x-ray images. For both the synchrotron-based and x-ray tube-based systems, imaging times of a few minutes were achieved. Scatter images acquired with the systems have contrast between pairs of materials such as plastics or muscle and fat much higher than conventional primary images. The achieved imaging times are similar to the duration of nuclear medicine procedures, demonstrating technical feasibility for clinical applications such as bone imaging and breast imaging.

7.2 Future work

7.2.1 Applications

The utility of the x-ray scatter imaging systems developed in this thesis is determined by how well they perform in different applications. Bone imaging is the application of most interest in the short term. Animal and human bone specimens would be imaged to determine the system's ability to assess bone health. Osteoporotic fracture risk is most often diagnosed using the patient's clinical history and bone mineral density (BMD) measure-

ments from dual energy x-ray absorptiometry (DXA).²¹² However, BMD accounts for only $\approx 70\%$ of the variation in bone strength²¹³ and nearly half of patients who suffer a fragility fracture have non-osteoporotic BMD values from DXA.²¹⁴ Other properties that need to be considered when predicting bone strength are bone mineral concentration, crystal size, chemistry, and microstructure,^{215,216} and coherent x-ray scatter measurements, which depend on the details of the chemical composition of the sample, are ideal for probing such characteristics.

Breast imaging is also of interest because fibroglandular tissues and tumours have low contrast in conventional transmission imaging but can be distinguished using their coherent scatter signatures.¹³¹ Furthermore, the maximum object thickness that the system can image encompasses that needed for mammography, however, patient imaging is not possible with the current prototype system.

Preliminary images of healthy canine prostate samples were acquired with the synchrotron-based system.¹⁸⁷ Structures not present in the primary images were visible in the scatter images, which may justify further investigation using diseased samples.

7.2.2 X-ray tube-based continuous motion acquisition

An advantage of the synchrotron-based system over the x-ray tube-based system is its ability to acquire scatter data with continuous object motion, which was shown to be about 30 times faster than step-and-shoot acquisition for the synchrotron system. It may be possible to do continuous motion acquisition with the x-ray tube-based system using the generator's fluoroscopy mode, for which a low current (up to 3 mA for our generator) is continuously applied to the x-ray tube.

Using all fifteen pencil beams, a $6.0\text{ cm} \times 10.0\text{ cm}$ region can be imaged with 1.0 mm pixel spacing by acquiring twenty 2.0 cm wide rows. The flat-panel detector would be continuously read out at its maximum frame rate giving 133 ms per scatter pattern, for

which the object would be translated at 7.5 mm s^{-1} so that it moves 1.0 mm during the time that one frame is acquired. At this speed, it would take 2.7 s to traverse one 2.0 cm row, which can be rounded up to 3 s to account for acceleration time and the time for vertical translation of 1.0 mm to the next row. The total acquisition time for twenty rows would therefore be about 60 s, nearly nine times faster than step-and-shoot for the x-ray tube-based system.

The mAs per scatter pattern would be 0.4 mAs for a tube current of 3 mA. The plastics image in Fig. 6.5 (a) was acquired with 1.7 mAs per scatter pattern, and the polyethylene-polystyrene SDNR was determined to be 6.4. Assuming the SDNR is proportional to the square root of the mAs at these exposure levels, the mAs can be lowered to about 1 mAs while maintaining $\text{SDNR} \geq 5$, and so image quality would likely be unacceptable for continuous motion fluoroscopic acquisition at 3 mA. Since the maximum cooling rate of the x-ray tube anode is 925 W, it should be possible to modify the generator to output a current of up to 11.3 mA for continuous fluoroscopic exposure at 110 kVp without exceeding the thermal limits of the anode, or even higher if exposures are restricted to less than 60 s and the tube is given time to cool afterwards. For a tube current of 10 mA, the mAs per scatter pattern would be 1.3 mAs, giving an image with lower SDNR than in Fig. 6.5 (a), but still expected to be acceptable.

For continuous motion acquisition, it would not be possible to acquire a dark frame in between each exposure frame. However, since the overall acquisition would be shorter and the power input would be lower, anode drift might not be a concern for fluoroscopic exposure, and interspersed dark acquisitions may not be necessary. The images would be affected by other artefacts, such as motion blur and possibly striping artefacts since the detector would be continuously read out while being irradiated.

7.2.3 Disentangling scatter patterns

The MLEM algorithm does not perfectly disentangle overlapping scatter patterns for multi-beam acquisition, giving ghosting artefacts, as seen in the scatter images in Figs. 4.2 and 4.3. The ghosting can be particularly severe if the Laue rings from the different beams have very different intensities. It is possible that machine learning techniques such as convolutional neural networks (CNN) that take a scatter pattern matrix as input and output a scatter profile vector for each beam can provide better results. The challenge would be providing enough data for training the neural network and it would likely be necessary to simulate scatter profiles. Initial studies with a CNN trained on five-beam scatter patterns generated from a bank of 1000 simulated scatter profiles showed that the CNN disentangled test simulated scatter patterns very well but when applied to real scatter patterns acquired with the synchrotron-based system the resulting scatter images were poorer than those reconstructed using MLEM. The performance of the CNN with the test scatter patterns suggests that its performance with real data might be improved by tuning the hyperparameters of the CNN, generating simulated scatter profiles that more closely resemble real scatter profiles, or acquiring a large bank of real scatter profiles to train with.

Alternatively, collimation can be added in between the object and detector to prevent scatter overlap altogether. Thin vanes made of W / Cu alloy extending from the flat-panel detector toward the object would restrict the scatter from each pencil beam to only reach a designated sub-area of the detector (Fig 7.1). Scatter image acquisitions were simulated¹⁸⁹ with straight vanes inserted between five pencil beams in a one-dimensional array. While there was no ghosting, the SDNR in images was reduced by a factor equal to the square root of the ratio of scatter-intensity-without-vanes to scatter-intensity-with-vanes, assuming the scatter profiles could be perfectly disentangled when not using vanes. Specifically, for straight vanes used with parallel beams having the same separation as in the synchrotron-based system, the SDNR was reduced by a factor of 1.54, which may be acceptable consid-

Figure 7.1: Thin vanes extending from the flat-panel detector would prevent overlap of scatter from different pencil beams.

erding the performance and computation time of the MLEM algorithm. With more beams and therefore less detector area per beam, more coherent scatter would be lost from each beam, decreasing the SDNR and the efficiency of the system. The loss can be mitigated by using curved vanes to more equally divide the detector area among the beams.

7.2.4 X-ray tube-based system primary imaging capability

X-ray scatter imaging systems utilize the additional information available in the coherently scattered photons that are always present when an object is irradiated but unused in conventional transmission imaging. A scatter imaging system should also be configured for optimal measurement of the always-present transmitted photons. For the synchrotron-based system, primary measurement was done by using discrete photodiodes and attaching a W / Cu blocker bar across the detector width to prevent scatter from the photodiodes and

their cables from reaching the flat-panel detector. Some of the coherent scatter from the object was also blocked by the bar.

The x-ray tube-based system has a two-dimensional array of fifteen pencil beams which would require three blocker bars to measure transmitted x rays with discrete photodiodes. It was decided that the area of the detector blocked by the bars would be unacceptable and instead attenuator discs were placed on the detector in the path of the pencil beams to lower their intensity to be within the dynamic range of the flat-panel detector. These discs over-attenuated the transmitted x rays and the primary images were practically unusable. The primary images could be improved by using thinner attenuators or by finding a way to use discrete photodiodes without disturbing the coherent scatter. One possibility if using pre-detector vanes (Sec. 7.2.3) is to incorporate into the vane structure the blockers for scatter from the photodiodes. If photodiodes are used, a scintillator that produces visible light photons in the range of wavelengths that the photodiode is sensitive to should be affixed to each photodiode to improve quantum detection efficiency.

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