Analyzer-based phase-contrast imaging system using a micro focus x-ray source

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Here we describe a new in-laboratory analyzer based phase contrast-imaging (ABI) instrument using a conventional X-ray tube source (CXS) aimed at bio-medical imaging applications. Phase contrast-imaging allows visualization of soft tissue details usually obscured in conventional X-ray imaging. The ABI system design and major features are described in detail. The key advantage of the presented system, over the few existing CXS ABI systems, is that it does not require high precision components, i.e., CXS, X-ray detector, and electro-mechanical components. To overcome a main problem introduced by these components, identified as temperature stability, the system components are kept at a constant temperature inside of three enclosures, thus minimizing the electrical and mechanical thermal drifts. This is achieved by using thermoelectric (Peltier) cooling/heating modules that are easy to control precisely. For CXS we utilized a microfocus X-ray source with tungsten (W) anode material. In addition the proposed system eliminates tungsten's multiple spectral lines by selecting monochromator crystal size appropriately therefore eliminating need for the costly mismatched, two-crystal monochromator. The system imaging was fine-tuned for tungsten $K\alpha_1$ line with the energy of 59.3 keV since it has been shown to be of great clinical significance by a number of researchers at synchrotron facilities. In this way a laboratory system that can be used for evaluating and quantifying tissue properties, initially explored at synchrotron facilities, would be of great interest to a larger research community. To demonstrate the imaging capability of our instrument we use a chicken thigh tissue sample. © 2014 AIP Publishing LLC. [<http://dx.doi.org/10.1063/1.4890281>]

I. INTRODUCTION

X-ray imaging is playing a leading role in modern medical imaging. Conventional X-ray imaging relies on absorption contrast, which is created when X-rays photons are removed from the beam by absorption or scattering. Therefore tissues with high attenuation and scattering properties show excellent visibility, e.g., bone tissue. However, the soft tissues like tendons, cartilages, muscle as well as tumors tissue are neither visible nor outlined because these tissues have similar attenuation properties.

Analyzer based phase contrast X-ray imaging (ABI),¹⁻⁵ schematically shown in Figure 1, is an emerging imaging technique that can increase soft tissue visibility by utilizing phase contrast.⁶ As the X-ray beam passes through the tissue, the beam intensity and phase change due to attenuation and refraction tissue properties. The alteration in beam phase is equivalent to that of the change in the beam propagation direction. For the most biological tissue this direction change is in the order of μrad .

In ABI raw images are acquired by measuring (sampling) the angular intensity profile (AIP) of an X-ray beam passing through a tissue at different angular positions (see Figure 1). This is achieved by using two crystals in the imaging system. The first crystal is used to obtain a quasi-monochromatic and collimated beam. The second one, called the analyzer crystal,

acts as an angular filter by reflecting only those components of the beam that are travelling at or near the Bragg angle of the crystal, thus rejecting all components outside a narrow angular range. By rotating (rocking) the analyzer crystal (in a short analyzer), it is possible to gather the information needed to compute the intensity and directionality characteristics of the transmitted beam. Angular intensity profile of the imaging system (no object present) is usually called rocking curve. The imaging beam at the location of the object is usually wide (i.e., size in x axis direction) but only several millimeters in the vertical height (i.e., size in y axis direction). Therefore the object is mounted on a scanning stage driven by a stepping motor, thus allowing object scanning in the vertical direction.

In the well-known diffraction enhanced imaging technique (DEI),¹ the raw image acquisition is done at two analyzer angular positions and it can simultaneously estimate two parametric images containing information about tissue's refraction and absorption.

As an additional benefit, since the contrast in ABI does not depend just on absorption, this imaging system can use a high-energy X-ray beam and, therefore, the radiation dose to the tissue (and in future developments to the patient) can be greatly decreased compared to traditional X-ray radiography. Keyriläinen⁷ showed that the ratio of absorption length and the phase shift length for typical tissue composition is dominated by phase, especially at higher energies.⁸

Finally, ABI parametric images, due to the narrow angular acceptance of the analyzer, are virtually immune to degradation due to Compton scatter at higher angles which

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FIG. 1. Schematic diagram of the experimental setup.

can help to eliminate breast compression as in conventional mammography.

For these reasons, ABI is emerging as a potential alternative to conventional radiography.^{9–15} As additional reading see Refs. 16 and 17 for a border overview of ABI and Refs. 18–23 for phase-sensitive imaging in general.

Synchrotron radiation has become the main tool for the phase contrast X-ray imaging and experimenting mainly due to its high intensity source that allows collimation and monochromatization of the beam while still having sufficient number of photons for imaging.

Few groups reported efforts to implement a benchtop system employing a conventional x-ray tube source (CXS).^{24–28} A number of obstacles are identified in CXS ABI systems. For instance, system alignment stability problems have been reported due to rotating anode vibration and system thermal variations^{25,27,28} or image degradation due to dispersive spread of virtual sources^{24,29} and undesired multiple spectral lines, $K\alpha_1$ and $K\alpha_2$, images.^{14,24,26,27} In addition, CXS are limited by thermal loading constraints of the anode material, and alternative tube configurations are under investigation.³⁰

Existing CXS ABI systems use: Molybdenum anode X-ray tube and imaging at $K\alpha_1$ energy of 17.48 keV,²⁴ stationary silver anode and imaging at $K\alpha_1$ of 22.16 keV,²⁷ stationary copper anode and imaging at $K\alpha_1$ energy of 8.05 keV,²⁵ stationary tungsten anode and imaging at $K\alpha_1$ energy of 59.3 keV,²⁶ and rotating tungsten anode at same energy.²⁸

The primary goal in the design of the experimental apparatus described here was to develop an ABI by using a CXS and imaging at hard X-ray energies (e.g., 59.3 keV). The key advantage of the presented system, over the few existing CXS ABI systems, is that it does not require high precision: electro-mechanical components, fixed anode CXS and X-ray detector. To overcome a main problem introduced by these components, identified as temperature stability, each component was kept at a constant temperature inside of an enclosure, thus minimizing its thermal drifts. This is achieved by using thermoelectric (Peltier) cooling/heating modules that are easy to control precisely. In addition the proposed system eliminates multiple spectral lines by selecting monochromator crystal size appropriately therefore eliminating need for the costly mismatched, two-crystal monochromator.²⁷

II. INSTRUMENT DESIGN

In this work we decided to build an imaging system using tungsten anode $K\alpha_1$ energy of 59.3 keV since it has been shown to be of great clinical significance.^{11,31} In this way a laboratory system used to advance evaluation and quantification of tissue properties, similar to Faulconer *et al.*,¹¹ Muehleman *et al.*,³¹ studies, would be of great interest to the larger research community.

A schematic diagram of the experimental setup is shown in Figure 1. The distance between each system component in the direction of beam propagation is minimized, in order to increase imaging beam flux, and usually determined by the size of motion controllers and the components' mounting assemblies. Next, Figure 2 shows a photograph of the actual setup.

FIG. 2. Experimental setup.

Some additional lead slits, shielding and safety interlock system, though important for a safe system operation, are not included in the schematic diagram since they are particularly setting depended.

In our system we used three enclosures: first encasing CXS; second one enclosing all electro-mechanical components including the stages, stage actuators, stage controllers, and both crystals; and third encasing the imaging detector. Each enclosure temperature was kept constant to minimize the thermal drift of electrical and mechanical parts within it. A special care was taken to thermally stabilize stages and controllers used for the analyzer crystals μrad angular movement. In Sec. II each subsystem is described in details.

A. Conventional X-ray source enclosure

A Kevex PXS10-16W microfocus tungsten X-ray tube was used as a radiation source. The maximum tube voltage and power are 130 kV and 16 W, respectively. The anode spot diameter is $21\text{ }\mu\text{m}$ at the maximum power, and only $5\text{ }\mu\text{m}$ at its lowest setting. Tungsten $K\alpha_1$ line with the energy 59.3 keV is used for imaging. An important aspect for this X-ray source is the small anode spot. In the proposed configuration the monochromator can only receive and diffract photons at or around $K\alpha_1$ energy coming only from a small source spot area. Therefore, using a microfocus source, a larger fraction of the emitted radiation can be utilized and the system efficiency is higher compared to larger focal spot systems. Detailed discussion of the monochromator crystal design will be given in Sec. III.

The system is extremely sensitive on crystal alignment and the anode spot y axis position (see Figure 1). Therefore we investigated anode spot drift with pin hole imaging methodology.³² The first sets of results are presented in Figure 3 which shows the anode spot movement, in x - y plane, from the moment the X-ray source was turned on. One can observe that after switching on the x-ray tube it takes around 20 min for the anode position to stabilize in the y-direction. The total measured drift was around $55\text{ }\mu\text{m}$. Additionally it took around 90 min to stabilize in the x -direction (direction not important for system alignment but may cause imaging blur) with the total drift around $25\text{ }\mu\text{m}$. After 90 min of operation the source *working temperature* was 29.5°C for 21°C ambient temperature. The second set of results, in Figure 4,

FIG. 3. Source anode spot movement after switching on the x-ray tube.

FIG. 4. Change of source spot position due to the ambient temperature variation.

shows the anode spot position deviation as the source working temperature deviates due to the ambient temperature variation. One can calculate that spot position, in y -direction, changes approximately $10.4\text{ }\mu\text{m}/^\circ\text{C}$ which translates to an unacceptable $61.4\text{ }\mu\text{rad}/^\circ\text{C}$ angular drift at the monochromator, for 169 mm distance between x-ray tube and monochromator (a distance denoted by l_1 in Figure 1).

These results indicate that to achieve a good system alignment and angular stability it is important to wait for the source to reach a stable working temperature. In addition it is essential to keep ambient temperature constant. We achieved this by enclosing the source and stabilizing enclosure temperature using TeTechnology, Inc Peltier Bi-directional, heating and cooling assembly. This method allowed us to achieve ambient temperature stability of $\pm 0.01^\circ\text{C}$ and focal spot stability of $\pm 0.1\text{ }\mu\text{m}$ which translated to $\pm 0.59\text{ }\mu\text{rad}$ angular drift. To reduce the cold start warm-up time, to 3–5 min, we kept CXS enclosure at 29.5°C when the source was off and at 21°C when on.

B. Electro-mechanical enclosure

The electro-mechanical enclosure contained the translational, and rotating stages for both monochromating and analyzing crystals as well as the y translational stage for the object. For example, the following electro-mechanical stages, manufactured by Thorlabs, are used: NRT150, TDC001, MTS50-Z8, NR360S, PZ630B, and PRM-1Z7.

1. Germanium (Ge) crystal monochromator stages

For the monochromator, a near perfect germanium crystal with etch pit density (EPD) lower than 500 cm^{-2} at (111) diffraction plane with *asymmetric cut* is used. For this crystal we measured about 54% reflectivity due to the impurities (note that perfect crystals with $\text{EPD} \sim 0$ should have 100% reflectivity). The monochromator was mounted on a linear travelling stage so that it could move along the y direction as denoted in Figure 1. The minimum step

size for the movement was $0.1\ \mu\text{m}$. The crystal size was $60\text{ mm} \times 10\text{ mm} \times 20\text{ mm}$. The crystal size in the x - y direction does not need specific consideration and can be chosen as available, however detailed consideration for a size in the z -direction and asymmetric cut selection will be given in Sec. III D.

2. Object stage

The object stage was mounted on a linear travelling and rotating stage so that it could move along the y direction and rotate around y axis. The linear stage is used for object vertical scanning with the minimal movement of $0.1\ \mu\text{m}$.

3. Ge crystal analyzer stages

The analyzer crystal is one of the most critical components in ABI phase contrast imaging, since it converts the X-ray phase change into intensity change due to the narrow and sharp shape of its rocking curve (AIP of the imaging system). A near perfect germanium crystal was used as an analyzer. It was obtained from the same ingot and sized as monochromator crystal, with EPD < 500 , and used at (111) diffraction plane with *symmetric cut*. The analyzer was mounted on a linear travelling stage, two rotational stages, and a piezoelectric stage. It could move along the y direction and rotate around the x axis. The minimum step size for linear movement was $0.1\ \mu\text{m}$, and for rotational stage was 0.0002° ($3.5\ \mu\text{rad}$). The piezoelectric stage was mounted on top of the rotational stages and it allowed fine-tuning of the crystal position. This stage has three piezoelectric actuators to precisely control the rocking angle (angle around x axis) and tilting angle (angle around z axis) of the analyzer. The angular resolution of the piezoelectric stage was $0.036\ \text{arc sec}$, or $0.17\ \mu\text{rad}$. Since piezoelectric actuators are controlled electrically, a special care was taken to thermally stabilize piezoelectric controllers. As before, we achieved this by enclosing the electromechanical components in an enclosure where the temperature was controlled using TeTechnology, Inc Peltier cooling assembly. This method allowed us to achieve temperature stability of $\pm 0.01^\circ\text{C}$ on the piezoelectric controller and $\pm 0.1^\circ\text{C}$ in the rest of the enclosure. We did not observe any measurable variation in beam intensity or position due to the enclosure temperature variation.

C. Camera enclosure

In this experimental setup we used a pixelated (imaging) X-ray camera, specifically Shad-o-Box 2048 from Teledyne Rad-icon Imaging Corp. The collected images were $2048\ \text{pixels} \times 1024\ \text{pixels}$ where each pixel size was of $48\ \mu\text{m} \times 48\ \mu\text{m}$. Camera used a $\sim 65\ \mu\text{m}$ thick Gadox scintillation material ($\text{Gd}_2\text{O}_2\text{:Tb}$; Kodak Min-R 2190;) that gave the camera's full-width at the half maximum (FWHM) of $\sim 100\ \mu\text{m}$. The camera sensor (RadEyeEV) was chilled down to -2°C (maximum experimentally achieved was -25°C) to decrease the dark current. This was achieved using a specially designed helium flushed enclosure cooled by using TeTech-

nology, Inc Peltier cooling assembly. This method allowed us to achieve camera's temperature stability of $\pm 0.01^\circ\text{C}$. This low temperature and level of stability allowed us to use long exposure times (order of days if needed) and reuse offset and gain correction images over a period of several months.

This camera and scintillation material had measured average sensitivity of $0.16\ 1/\text{photon}$ at $60\ \text{keV}$ (one in every 6 photons produces a change at least significant bit). This measurement agrees with the absorption ability of $\sim 65\ \mu\text{m}$ thick Gadox.

D. X-ray spectrometer

The second detector in use was the XR-100T CdTe X-ray detector from Amptek Inc., which was used for system alignment. The effective detecting area was $3\text{ mm} \times 3\text{ mm}$. The energy resolution, according the manufacturer specifications, was $0.600\ \text{keV}$ @ $59.54\ \text{keV}$ measured using ^{241}Am source. This resolution was sufficient to separate tungsten $K\alpha_1$ ($59.3\ \text{keV}$) and $K\alpha_2$ ($57.982\ \text{keV}$) lines during alignment process.

Both the Amptek X-ray detector and the Shad-o-Box X-ray camera were mounted on two linear stages. They could be moved along the x and y directions so that they could be moved into the beam alternatively. The minimum step size for both linear stages was $0.1\ \mu\text{m}$.

III. DESIGN OF THE MONOCHROMATOR CRYSTAL

To design the monochromator, several crucial factors should be taken into account: (1) the monochromator angular acceptance should be increased to accept as much radiation as possible from the CXS anode and (2) the crystal size along z direction should be limited to diffract $K\alpha_1$ line only and no $K\alpha_2$ line should be seen in the diffracted beam. We will also briefly discuss the secondary effects introduced by choosing an asymmetrically cut monochromator, such as: increase in energy range acceptance and radiation dose as well as the impact on the final system resolution. As a final point, we will also discuss needed analyzer crystal size.

A. The monochromator angular acceptance

The natural Darwin width of Ge (111) crystal at tungsten $K\alpha_1$ energy is³³

$$w_0 = 2.218\ \text{arc sec} (10.75\ \mu\text{rad}). \quad (1)$$

When the Ge crystal is perfectly aligned, as shown in Figure 5, the ray hitting the crystal center C from the anode spot center O satisfies Bragg's law. The rays emitting from the other part of the anode spot can be diffracted at point C if they hit the crystal at an angle within Darwin width: $[\theta_B - \frac{w_0}{2}, \theta_B + \frac{w_0}{2}]$. So the crystal accepts radiation only from a h part of the anode which is

$$h_{1,2} = l_1 \tan\left(\theta_B \pm \frac{w_0}{2}\right), \quad (2)$$

$$\Delta h = h_1 - h_2 = 1.818\ \mu\text{m}, \quad (3)$$

FIG. 6. Diagram for calculating diffracted energies of the asymmetric cut Ge crystal.

Now our design objective is to choose the crystal size so that $\theta_2 < \theta_c = 1.875^\circ$, that is, θ_2 should be less than the Bragg angle for $K\alpha_2$ at 57.982 keV. After some algebra we can arrive at the following expression:

$$u_{\max} = \frac{l_1(\theta_c - \tan(\theta_B)) - \frac{S}{2}}{(\theta_c - \tan(\alpha))} = 2.527 \text{ cm} > u. \quad (16)$$

As for the minimal crystal width one can show that

$$u > u_{\min} = \frac{S}{2} \left(\frac{1}{\tan(\theta_B) - \tan(\alpha)} \right) = 0.305 \text{ cm} \quad (17)$$

would ensure that any $K\alpha_1$ energy photon leaving the source at a Bragg angle would be diffracted by the monochromator.

For our system we choose $u = 1 \text{ cm}$ leaving a large margin on both requirements. This defined the total monochromating crystal length of $2u = 2 \text{ cm}$.

To experimentally verify this theoretical consideration we measured (shown in Figure 7) beam spectrum before and after monochromator. One can clearly identify $K\alpha_1$ and $K\alpha_2$ lines in the beam before monochromator while no $K\alpha_2$ line is visible in beam after.

E. Analyzer crystal size

Considering a case where all of $K\alpha_1$ radiation after the monochromator is diffracted from the analyzer, and the fact that we are dealing with a fan beam, one can show that the

FIG. 7. Beam spectrum of X-ray source: after monochromator and after analyzer.

FIG. 8. Pictures of the chicken leg bone used as the sample in this work, showing the front side (left) and back side (right).

analyzer crystal minimal width should be^{29,36}

$$u_2 > u_{\min} \frac{(|\frac{1}{b}|l_1 + l_2 + l_3)}{|\frac{1}{b}|l_1} = u_{\min} \cdot 1.097 = 0.329 \text{ cm}. \quad (18)$$

For our system we choose the same crystal size as for a monochromator so $u_2 = 1 \text{ cm}$.

From experimental results of beam spectrums after monochromator and after analyzer crystal (shown in Figure 7), one can identify that beam after monochromator at $K\alpha_1$ line has FWHM of 0.951 keV (with peak at $8.42 \times 10^5 \text{ ph/cm}^2 \text{ s keV}$) and beam after analyzer crystal FWHM of 0.697 keV (with peak at $1.97 \times 10^5 \text{ ph/cm}^2 \text{ s keV}$). Note that the spectrometer resolution is 0.6 keV. The reduction of the flux after monochromator to flux after analyzer can be calculated to be $4.27 \times$. We attributed analyzer energy acceptance of ~ 2.26 , as calculated in consideration after Eq. (10), and crystal reflectivity of $\sim 54\%$ since crystals in use are not perfect and have $\text{EPD} < 500$. The measured in beam flux post monochromator was $8.0 \times 10^5 \text{ ph/cm}^2 \text{ s}$ and a beam height of 0.33 mm.

IV. IMAGE ACQUISITION AND PROCESSING

A. Imaging acquisition software

The image acquisition software is developed for MATLAB and Active X environment. The software allows users to control all system components.

B. Image acquisition

The DEI algorithm computes parametric images of the refraction angle and the attenuated intensity from two “raw” images measured at two rotational positions of the analyzer crystal (see Figure 1). In DEI algorithm the two positions are chosen to be on opposite sides of the AIP (rocking curve) of the imaging system. Specifically, images are taken with the analyzer positioned at $5 \mu\text{rad}$ less or greater than the peak intensity of the rocking curve. Raw images shown in Figure 9 are obtained by imaging a piece of chicken leg bone attached with skin, muscles, tendons, and cartilages, as shown in Figure 8. The height of the sample was 20 mm.

Since the height of the beam irradiating the sample was around 0.5 mm, the tissue sample was imaged several pixel lines at the time. A linear travelling stage was used to move the tissue along the y direction. The step size for the sample movement was set at a value so that the change of image position on the camera for each step was an integer in pixels,

FIG. 9. Raw images obtained by the presented DEI system for the chicken leg tissue sample. Images are taken with the analyzer positioned at 5 μ rad less or greater than the peak intensity of the rocking curve.

considering the vertical beam divergence. The step size for moving the sample along the y direction was 0.0909 mm. So the image moved 0.15 mm, or 3 pixels on the camera for each step. 230 steps were used for a whole image. The exposure time for each image was 300 s.

The beam intensity was monitored at each step. An automatic realignment procedure for the monochromator and an-

FIG. 10. Calculated DEI parametric images for the chicken leg tissue sample. Top absorption image. Bottom refraction image.

alyzer crystals was done when the change in the beam maximum intensity exceeded 5%. The total collecting time for one image was around 40 h, of which only 1.5 h was lost on automatic realignment.

Total radiation dose delivered to our sample was 2.1 mGy.

C. Image processing

The measured raw images are stitched together, taking into account the amount and direction of the sample movement. The stitched raw images are showed in Figure 9. Next, in Figure 10 we show absorption and refraction images calculated using the DEI.¹ One can note all the details visible in these images, particularly the cartilage layer at the left end of the bone (marked by white arrow in Figure 9). The utilities of absorption and refraction images are explored in many publications, for summary see Refs. 16 and 17.

V. CONCLUSION

A design for an analyzer based phase contrast X-ray imaging, using a conventional x-ray source, was presented and successfully used for DEI imaging. Experimental data confirmed that the system provides great soft tissue contrast even though a high energy X-ray beam was used. The presented instrument avoids using high precision components which is compensated by keeping the system components at a constant temperature inside of three enclosures. In addition the system eliminates multiple spectra lines by selecting monochromator crystal size correctly therefore eliminating need for the costly mismatched, two-crystal monochromator.

One problem of this imaging system is the long acquisition time. We envision this can be reduced by at least a factor of ~ 6 with use of a better imaging camera designed specifically for high-energy photons. In addition to reduction in imaging time this would also reduce radiation dose, without loss of image quality, to 0.35 mGy. The use of a conventional x-ray source and conventional electro-mechanical components offers the possibility to bring this technique into research laboratories.

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