Three Energy Computed Tomography with Synchrotron Radiation

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THREE ENERGY COMPUTED TOMOGRAPHY WITH SYNCHROTRON RADIATION

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Preliminary experiments for digital subtraction computed tomography (CT) at the K-edge of iodine (33.17 keV) were carried out at SMERF (Synchrotron Medical Research Facility X17B2) at the National Synchrotron Light Source, Brookhaven National Laboratory. The major goal was to evaluate the availability of this kind of imaging for invivo neurological studies. Using the transvenous coronary angiography system, CT images of various samples and phantoms were taken simultaneously at two slightly different energies bracketing the K-absorption edge of iodine. The logarithmic subtraction of the two images resulted in the contrast enhancement of iodine filled structures. An additional CT image was taken at 99.57 keV (second harmonic of the fundamental wave). The third energy allowed the calculation of absolute iodine, tissue and bone images by means of a matrix inversion. A spatial resolution of 0.8 LP/mm was measured in single energy images and iodine concentrations down to 0.082 mg/ml in a 1/4" diameter detail were visible in the reconstructed subtraction image.
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1. Introduction

Monochromatic computed tomography (CT) projects for medical imaging are underway at the NSLS (National Synchrotron Light Source) at Brookhaven National Laboratory, Upton, NY and at the ESRF (European Synchrotron Radiation Facility), Grenoble, France. Imaging techniques being developed include K-edge subtraction tomography at the K-edge of contrast agents such as iodine, gadolinium and xenon, as well as dual energy photon absorptiometry (DPA-CT). For the latter a low energy monochromatic CT image (\( E_\gamma \approx 40 \) keV) and a high energy image (\( E_\gamma \approx 80 \) keV) are obtained and subtracted.

The use of monochromatic X-rays is expected to yield better contrast than in a conventional CT system for a given dose and spatial resolution eliminating beam hardening effects. The K-edge subtraction images will allow quantitative concentration measurements of the contrast agent. DPA-CT will give a map of medium Z elements. The primary applications will be in brain research measuring blood flow and tissue perfusion measurements for studies of cerebral ischemia and brain tumors as well as plaque composition studies in carotid arteries.

This paper presents preliminary results obtained on phantoms and anatomical samples using the K-edge monochromatic imaging system built at the NSLS for transvenous coronary angiography. Neither institute has a complete clinical system available yet.

2. Materials and Methods

The NSLS X17B2 beamline uses hard X-rays from a superconducting wiggler installed on the 2.5 GeV electron storage ring [1-3]. Two beams bracketing the iodine K-edge (33.17 keV) are produced using a single crystal bent Laue monochromator and a splitter [4,5]. This results in two X-ray beams 15 cm wide vertically separated initially but cross at the patient position during
coronary angiography imaging. The patient is normally scanned vertically through the beams and both energy images are obtained line by line [figure 1]. Both beams are detected simultaneously on a dual line multi element LiF drifted Si detector. For the detector used [6], each line consists of 300 pixels with a pitch of 0.5 mm. The imaging system includes a patient safety system and dose monitoring equipment.

To convert the angiography system to a K-edge tomography setup adequate for phantom imaging, it was sufficient to remove the patient chair and install a rotation stage on the existing positioning system. The rotation axis had to be set normal to the mean plane of the X-ray beams at the crossing point. The angle between the two beams was considered to have a negligible effect on the reconstructed difference image.

The phantoms imaged with the dual energy system included a spatial resolution phantom and a contrast phantom. In addition a biological sample was imaged using three energies. The spatial resolution phantom was a cylinder of Plexiglas of diameter 80 mm with a hole pattern as shown in figure 2. The contrast phantom (see figure 3) consisted of a 100 mm diameter cylinder of Plexiglas with holes of various sizes filled with iodine solutions of varying concentrations. This phantom was imaged a number of times with different iodine concentration patterns. A spine segment of a cow tail surrounded by tissue with two plastic tubes filled with iodine solution inserted in the tissue was imaged below and above the K-edge simultaneously. A third image was taken at approximately 99 keV by using the second harmonic radiation diffracted by the monochromator. The primary beams at 33 keV were attenuated with a tin filter.
3. Results

Figure 2 shows a portion of the low energy image of the spatial resolution phantom. We can clearly distinguish the 0.5 mm holes which are also separated by 0.5 mm. The spatial frequency spectrum of a profile taken through the line of 0.5 mm holes shows that the spatial resolution of the system is 0.8 LP / mm on the reconstructed image, which is consistent with the pixel size. Figure 3 shows the iodine image of our contrast phantom for the lowest concentration visible. Dose was not precisely recorded but was of the same order of magnitude as that used in a clinical conventional CT scanner for brain examinations. The lowest concentration visible is on the order of 80μg/ cm³. The measured concentration on the reconstructed image was 76μg/ cm³ and the difference with the nominal concentration of 82 μg/ cm³ is not significant. The circular artifacts visible on the image are due to noisy channels in the detector.

In figure 4 the top three images are respectively the reconstructed images below and above the iodine K-edge and at 99 keV. The bottom three images are “subtraction” images. They were obtained using the mass attenuation coefficient matrix given in equation 1.

\[
\begin{pmatrix}
S_1 \\
S_2 \\
S_3
\end{pmatrix} =
\begin{pmatrix}
\mu_{\text{iodine}}^1 & \mu_{\text{water}}^1 & \mu_{\text{bone}}^1 \\
\rho_{\text{iodine}} & \rho_{\text{water}} & \rho_{\text{bone}} \\
\mu_{\text{iodine}}^2 & \mu_{\text{water}}^2 & \mu_{\text{bone}}^2 \\
\rho_{\text{iodine}} & \rho_{\text{water}} & \rho_{\text{bone}} \\
\mu_{\text{iodine}}^3 & \mu_{\text{water}}^3 & \mu_{\text{bone}}^3 \\
\rho_{\text{iodine}} & \rho_{\text{water}} & \rho_{\text{bone}}
\end{pmatrix}
\begin{pmatrix}
\rho_{\text{iodine}} & 1 - \text{energy} = 33.019 \text{ keV} \\
\rho_{\text{water}} & 2 - \text{energy} = 33.319 \text{ keV} \\
\rho_{\text{bone}} & 3 - \text{energy} = 99.957 \text{ keV}
\end{pmatrix}
\]

In equation (1), S denotes the vector of the normalized sinograms measured at the three energies, t the transmission path through the specific material and ρ and μ are the densities and the absorption coefficients of water (tissue), bone and iodine [7]. The inversion was performed on the sinograms before reconstruction. The reconstructed images are therefore, the tissue map, bone map and
iodine map from left to right in the lower line of figure 4. The artifacts in the 99 keV image are due to the low photon flux available at that energy.

4. Conclusion

An optimized clinical system requires all the monochromatic X-ray beams to follow the same path through the object or organ to be imaged. This implies that a clinical monochromatic CT scanner can be very similar to the coronary angiography system on beamline X17B2 at NSLS but with a fixed exit monochromator. Otherwise line scan radiography and tomography are extremely similar imaging procedures. The preliminary images taken during the experiment reported here confirmed the design options considered for a clinical monochromatic CT scanner for brain studies. [9]

5. Acknowledgments

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6. References

Figure Captions

Figure 1.: Sketch of the SMERF beamline for transvenous coronary angiography at the NSLS at BNL, Upton, NY in a possible CT set up.

Figure 2.: Part of the low energy image of the spatial resolution phantom and profiles taken at the indicated positions.

Figure 3.: Iodine image of the contrast phantom for a nominal concentration of 0.082 mg /ml. The positions of the holes is shown in the sketch above.

Figure 4.: The three images upper row are the reconstructed images below, above the iodine K-edge and at 99 keV. The bottom images are tissue, bone and iodine images obtained using the inversion of the mass attenuation coefficient matrix.
The diameter of holes is equal to the spacing.

FFT from the profiles delivers the spatial frequency which can be interpreted as the spatial resolution.
$0.082 \text{ mg/ml iodine}$

measured concentrations

$0.076 \pm 0.01 \text{ mg/ml}$ in reconstruction
low energy image 33.019 keV  medium energy 33.319 keV  high energy 99.957 keV

tissue image  bone image  iodine image (1mm tubing filled with 50 mg/ml iodine is clearly visible)