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## COMPUTED TOMOGRAPHY USING SYNCHROTRON RADIATION

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### ABSTRACT

X-ray computed tomography (CT) is a widely used method of obtaining cross-sectional views of objects. The high intensity, natural collimation, monochromaticity and energy tunability of synchrotron x-ray sources could potentially be used to provide CT images of improved quality. The advantages of these systems would be that images could be produced more rapidly with better spatial resolution and reduced beam artifacts. In addition, images, in some cases, could be acquired with elemental sensitivity. As a demonstration of the capability of such a system, CT images were obtained of four slices of an excised pig heart in which the arteries and the cardiac chambers were filled with an iodinated medium. Images were taken with incident x-rays tuned successively to energies just above and below the iodine K edge. Iodine specific images were obtained by logarithmically subtracting the low energy image data from the high energy data and then reconstructing the image. CT imaging using synchrotron radiation may become a convenient and non-destructive method of imaging samples difficult to study by other methods.

## INTRODUCTION

X-ray computed tomography (CT) is a widely used method of obtaining cross-sectional views of objects. This field was pioneered by Hounsfield [1] in 1973 and now many commercial systems, using special high voltage x-ray tubes are available, primarily for medical imaging. Synchrotron radiation sources potentially could provide a better source of x-rays since they produce very intense, naturally collimated, narrow bandwidth and tunable photon beams.

Synchrotron sources offer several advantages for CT imaging. Principally, by taking two images above and below an x-ray absorption edge, it is possible to reconstruct the spatial distribution of a specific element. In addition, the higher intensity could allow real-time imaging of moving samples, providing the samples could withstand rapid rotation. Beam hardening artifacts will not be present as the photons are almost monochromatic. The background signal due to scattered photons is reduced since the beam has a divergence of only several mrad and the transmitted radiation can be collimated so that the detector sees very little scattered radiation. Radiation damage to the sample can be minimized by tuning the beam energy to give the maximum signal-to-noise ratio. A final advantage is that small samples could be studied with high spatial resolution in reasonable sampling times.

The application of synchrotron radiation to CT has been studied theoretically by Grodzins [2] who found that water-like samples of  $100 \mu\text{m}^2$  size with a resolution element size of  $1 \mu\text{m}^2$  could be studied with a sensitivity below 50 ppm for elements from phosphorus to bromine. Many types of samples could be examined and elemental images could be reconstructed providing the samples were thin enough to transmit enough x-rays near an appropriate absorption edge. Since the sample must be rotated, it may not be possible to use synchrotron sources for CT imaging of human subjects.

As a demonstration of the capability of a synchrotron CT system, trial images were obtained of four slices of an excised pig heart in which the arterial tree and the cardiac chambers were filled with iodinated vaseline. Images were taken with incident x-rays tuned successively to energies just above and below the iodine K edge. Iodine specific images were obtained by logarithmically subtracting the low energy data from the high energy data and then reconstructing the image.

## DATA ACQUISITION PROCEDURE

The x-ray tomographic method requires the measurement of the x-ray transmission coefficients through each resolution element at a large number of angles. Usually the resolution element size is in the range  $1$  to  $20 \text{ mm}^2$  and more than 100 different angular measurements are made. Using this data an image is then reconstructed of the slice through the sample. There are a variety of algorithms which perform the reconstruction at high speed with modern minicomputers.

The method used in this study is shown schematically in Fig. 1. The white radiation synchrotron beam is monochromatized, vertically collimated, passed through the sample and measured with a horizontal one-dimensional detector.

Specifically, a 2 cm wide beam from a wiggler insertion device at the Stanford Synchrotron Radiation Laboratory (SSRL) was monochromatized by a standard SSRL double crystal monochromator to produce a beam with a fractional energy spread of about  $5 \times 10^{-4}$ . The energy of the beam was selected by angular adjustment of the monochromator. The vertical height of the beam was collimated to 0.5 mm and the total beam intensity monitored with a standard SSRL argon-filled ionization chamber. The beam was then passed through the sample mounted on a computer controlled rotary stage. Following the sample was a one-dimensional multi-element detector which measured the beam intensity in horizontal segments of 0.5 mm. This detector, which has been described previously by Thompson et al [3], was a cooled (-30 deg C) multi-element lithium-drifted silicon detector with a thickness of 5 mm. The x-ray

flux in each element was measured by connecting a preamplifier and voltage-to-frequency converter to each element, as described by Zeman et al [4], and measuring the frequency with a computer. This system has excellent dynamic range, linearity and stability. Since resolution elements of 0.5 mm by 0.5 mm were desired, but the center-to-center spacing of the detector elements was only 1 mm, a comb slit was placed ahead of the detector and two interlaced sub-images taken. In addition, since the beam width was only 2 cm and the field-of-view necessary to cover the complete excised pig heart was 12 cm, the sample was translated in six steps of 2 cm and the data acquired in swaths which were combined in correct sequence before the image was reconstructed.

A CT image was obtained by recording the intensity in each of the detector elements at different angular positions of the rotary sample stage. The intensities were normalized using the ionization chamber ahead of the sample. Elementally sensitive images were made by taking two sets of measurements, one just below an absorption edge of the element to be measured and the other just above the absorption edge. The elemental image was obtained by subtracting the normalized logarithmic intensity at each point. For iodine-specific images one set of data was taken 25 eV below the iodine K edge at 33.169 keV and another set 25 eV above the edge.

The CT images were created using a Shepp-Logan[5] filter and back-projecting with linear interpolation. Synchrotron x-ray CT data has excellent statistics and difference images, in particular, have high contrast and sharp edges. Therefore, it is possible that the Shepp-Logan filter is not optimal and that a filter with a higher frequency cutoff would improve the reconstructed images.

## EXPERIMENTAL RESULTS

The experiment was done during March, 1982 with the SSRL storage ring operating at 3 GeV and 70 mA. The beam intensity was about  $4 \times 10^9$  photons/sec/mm<sup>2</sup> at 33 keV. The detector and imaging system was identical to that used in an evaluation of the use of synchrotron radiation for non-invasive angiography. Results with this system have been presented by Hughes et al [6].

An embalmed pig heart was used as a sample for this study. Using appropriate mixtures of ethiodol (an iodine containing compound) and vaseline, the arterial tree, the left and right ventricles, and the right atrium were filled. The mixture was adjusted to make the iodine concentrations in the arteries 20 mg/ml, in the left ventricle 20 mg/ml, in the left atrium 13 mg/ml and in the right ventricle and right atrium 7 mg/ml.

Two-dimensional x-ray pictures taken at incident photon energies 25 eV above and below the iodine K edge at 33.17 keV were first acquired at a variety of angles. The digitally subtracted logarithmic difference images were then calculated. A set of these images is shown in Figure 2. Fig. 2(a) is the low energy image, Fig 2(b) is the high energy image, Fig. 2(c) is the difference image and Fig. 2(d) is the difference image with the heart rotated 90 degrees. The four levels at which subsequent CT images were taken are marked by the numbered lines.

Normalized x-ray transmission data for the CT images were collected across a 12 cm width (240 resolution elements) and at angular steps of 1.8 deg (100 projections). Four sets of data were acquired at different heights on the heart. The second set of reconstructed images taken above and below the iodine K edge at position 2 are given in Fig. 3(a-b). The four difference images are presented in Fig. 4(a-d).

As can be seen very good images were obtained using synchrotron radiation. Although there are slight differences between the high and low energy images, the arteries are not visible in these images. However, when the difference images are calculated the arteries and ventricles stand out with excellent contrast. It is even possible to see that the larger arteries in these images are oblong since they are crossing the projection plane at an angle.

## CONCLUSIONS

CT images have been produced using a synchrotron x-ray beam line at SSRL. They demonstrate the ability to obtain both standard CT images and images which show the spatial distribution of only a specific element. When the capacity of tomographic imaging to produce cross-sectional images non-destructively is combined with a powerful synchrotron x-ray source a technique is provided to image samples which can not be studied with current x-ray sources. The high intensity, natural collimation, monochromaticity and tunability provide an almost ideal source for x-ray tomography. It can both improve CT imaging of samples by increasing the signal-to-noise ratio and also provide elemental sensitivity for many samples when the sample is thin enough to transmit x-rays near an appropriate absorption edge.

## ACKNOWLEDGEMENTS

The assistance of J. Stone in preparation of the pig heart samples is appreciated. The cooperation of the SSRL staff in the setup and operation of this experiment is gratefully acknowledged. This work was supported in part by the Director's Office of Energy Research, Office of Health and Environmental Research, U.S. Department of Energy under Contract No. DE-AC03-76SF00093, the Henry J. Kaiser Family Foundation and the National Science Foundation under contract DMR 77-27489 (for the operation of SSRL).

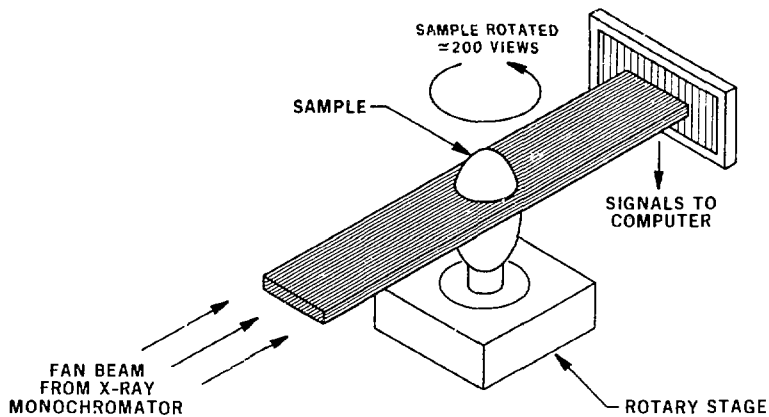
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## FIGURE CAPTIONS

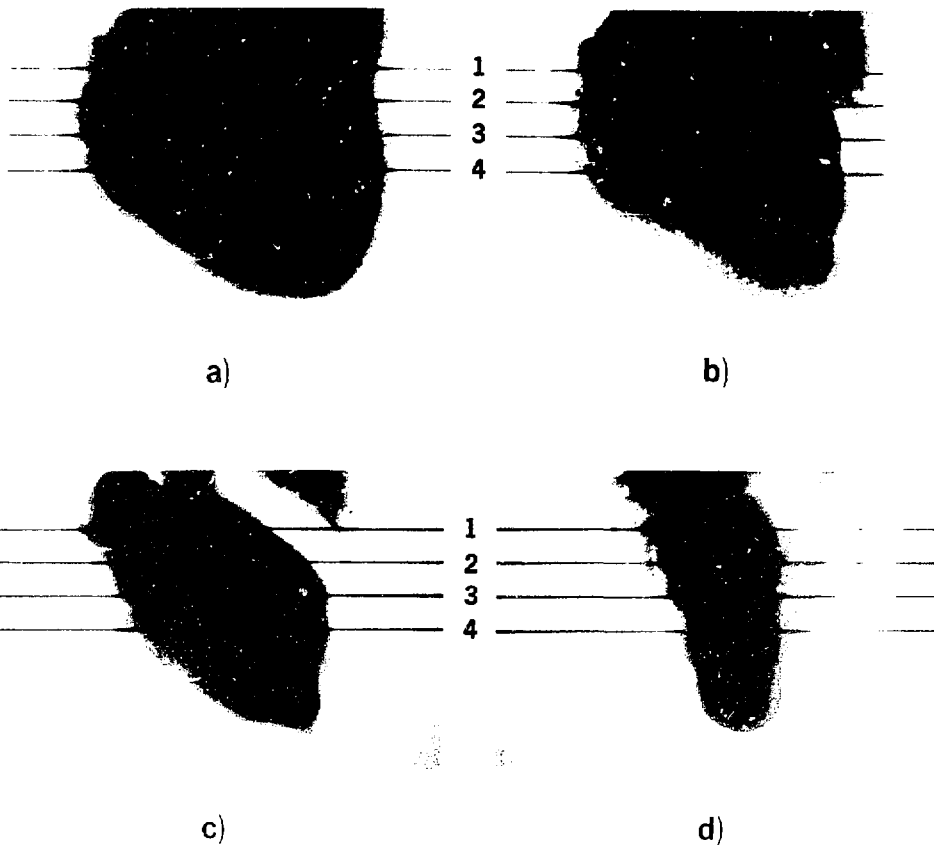
1. Schematic layout of a computed tomographic imaging system using synchrotron radiation.
2. Two-dimensional x-ray images of an embalmed pig heart. Fig. 2(a) is an image taken at 25 eV below the iodine K edge, Fig. 2(b) is the same image taken at 25 eV above the edge and Fig 2(c) is the difference image taken at the same position. Fig. 2(d) is the difference image taken with the pig heart rotated 90 degrees to Fig. 2(c). The four positions at which CT images were obtained is indicated by the numbers 1-4.
3. CT images obtained at incident photon energies 25 eV below (a) and above (b) the iodine K edge at vertical position 2 of the heart.
4. CT difference images obtained by logarithmically subtracting the low energy image data from the high energy image data before image reconstruction. The four images correspond to the four positions indicated in Fig 2.

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Fig. 1. Schematic layout of a computed tomographic imaging system using synchrotron radiation.



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Fig. 2. Two-dimensional x-ray images of an embalmed pig heart. Figure 2(a) is an image taken at 25 eV below the iodine K edge, Fig. 2(b) is the same image taken at 25 eV above the edge and Fig. 2(c) is the difference image taken at the same position. Figure 2(d) is the difference image taken with the pig heart rotated 90 degrees to Fig. 2(c). The four positions at which CT images were obtained is indicated by the numbers 1-4.





**a)**



**b)**

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Fig. 3. CT images obtained at incident photon energies 25 eV below (a) and above (b) the iodine K edge at vertical position 2 of the heart.



a)



b)



c)



d)

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Fig. 4. CT difference images obtained by logarithmically subtracting the low energy image data from the high energy image data before image reconstruction. The four images correspond to the four positions indicated in Fig. 2.

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