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MULTIPLE ENERGY COMPUTED TOMOGRAPHY WITH MONOCHROMATIC X RAYS FROM THE NSLS

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Abstract

We used monochromatic x rays from the X17 superconducting wiggler beamline at the National Synchrotron Light Source (NSLS), Brookhaven National Laboratory, for dual-energy quantitative computed tomography (CT) of a 27 mm-diameter phantom containing solutions of different KOH concentrations in cylindrical holes of 5-mm diameter. The CT configuration was a fixed horizontal fan-shaped beam of 1.5 mm height and 30 mm width, and a subject rotating around a vertical axis. The transmitted x rays were detected by a linear-array Si(Li) detector with 120 elements of 0.25 mm width each. We used a two-crystal Bragg-Bragg fixed-exit monochromator with Si<220> crystals. Dual photon absorptiometry (DPA) CT data were taken at 20 and 38 keV. The reconstructed phantom images show the potential of the system for quantitative CT.

I. INTRODUCTION

This paper reports the first phantom studies carried out with a prototype monochromatic CT system being developed at the X17 superconducting-wiggler at the NSLS [1]. The project, which has the goal of human brain CT imaging with monochromatic x rays, is called Multiple Energy Computed Tomography (MECT) [2,3].

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Monochromatic x rays provide better CT image contrast and better CT image quantitative precision than does wide-band bremsstrahlung radiation. Synchrotron radiation is a suitable and at present the sole source of monochromatic x rays for CT imaging. Use of a well-defined beam energy eliminates beam hardening effects [4], and allows implementation of improved dual-energy CT methods, like DPA CT [5] and K-edge subtraction (KES) CT with contrast elements such as iodine [6]. In particular DPA CT will enhance our ability to detect variations in the concentrations of intermediate-Z elements (i.e., P, S, Cl, K, Ca, and Fe) in the brain, some of which (such as K and Ca) have particularly important neurological significance .

II. EXPERIMENTAL METHOD

A. *The Imaging System*

We used a two crystal (Bragg-Bragg) monochromator with Si<220> crystals [7]. The Golovchenko design incorporates a coupled rotation and translation to the second crystal to hold them parallel, and to provide a fixed exit beam height [8]. The first crystal was mounted on a water cooled copper block with a In-Ga liquid metal as a thermal interface to stand the high heat load from the X17 wiggler, and the second crystal was detuned using a piezoelectric transducer in order to reject higher order harmonics [7]. The transducer's high voltage was controlled by a feedback system.

A cooled (-20°C) linear-array Si(Li) detector having 120 elements of 0.25 mm width and 3 mm depth [9] was used to detect the transmitted radiation.

We used a multichannel current digitizer that utilizes voltage-to-frequency convertors [10]. It has a dynamic range of 40,000:1 and a non-linearity of 0.02%; it is currently used in the Coronary Angiography Project at the NSLS [11].

III. MEASUREMENT RESULTS

Our phantom was a 27 mm-diameter plexiglas cylinder with 14 cylindrical, 5 mm-diameter axial holes. We used solutions of KOH, with 200, 37.7, 7.7, 1.5 or 0.0 mmolar concentration in the holes (see Fig. 1). The phantom was rotated 360° in 24 seconds. Projections were taken on the fly in a continuous mode at 0.25° intervals, i.e. every 1/60 second. The measurements were carried out at 20 and 38 keV.

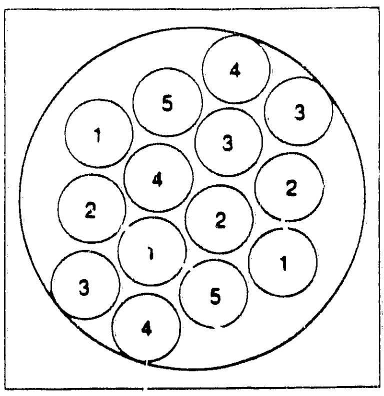

Fig. 1. Pattern of the holes in the phantom. The KOH concentration in the holes are 200 mmolar for hole 1, 37.7 mmolar for hole 2, 7.7 mmolar for hole 3, 1.5 mmolar for hole 4 and 0.0 mmolar in hole 5.

The following steps were followed in the data analysis:

1. Image corrections were applied to the 20-keV and the 38-keV images before and after tomographic reconstruction to reduce artifacts induced by bad detector channels (mostly in the central part of the detector) and other experimental imperfections. The results are shown in Figs 2 and 3, respectively. Because of the unresolved artifacts left in the central part of the images, the quantitative image analysis

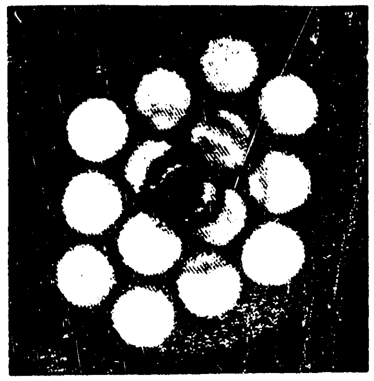

Fig. 2. Reconstructed image of the phantom at 20 keV beam energy.

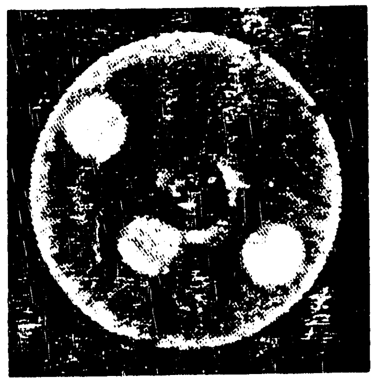

Fig. 3. Reconstructed image of the phantom at 38 keV beam energy.

steps described below were applied only to the 10 non-central holes.

2. For each image, the attenuation coefficient of the pixels in different holes (in units of 1/cm) were averaged over pixels in each hole and over holes of the same K concentration. The results were plotted as a function of the solution K concentration (see Figs 4 and 5). The slope of the linear fit provided the mass attenuation coefficient of K for the two nominal beam energies of 20 and 38 keV, while the intercept with the vertical axis provided that for water.

3. Using these K attenuation coefficients, the "effective" beam energy for these two energies were found to be about 22 and 39 keV. The effect was attributed to the harmonic contamination of the beam.

4. Using the measured attenuation coefficients, the dual photon absorptiometry formalism [12] was used to obtain separate images of the low-Z and the

intermediate-Z images. The results are shown in Figs. 6 and 7. Figs. 8 and 9 show the linear fit to the measured gray scale of the intermediate-Z and low-Z elements of Figs. 6 and 7, respectively, as a function of the K-concentration in the phantom holes.

IV. CONCLUSIONS

Despite of the image artifacts, the holes with 200, 37.7, 7.7 mM KOH concentration show decreasing gray scales. A K concentration of about 7.7 mM (0.3 mg/cc) is about 1/10 of the potassium concentration in the normal brain tissue. Besides that, the lack of dependence of the low-Z gray scale on K concentration (Fig. 9) shows the potential of DPA to separate these two groups of elements.

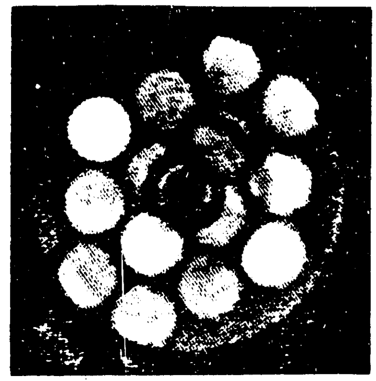

Fig. 6. Image of the intermediate-Z element group, obtained by applying the DPA formalism to the 20-keV and the 38-keV images from Figs. 2 and 3.

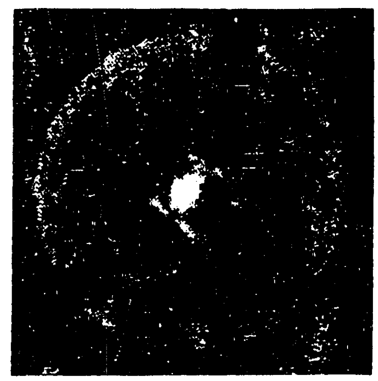

Fig. 7. The same as Fig. 6, except for the low-Z elements.

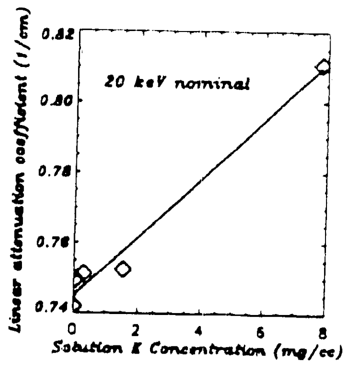

Fig. 4. Least-square fit of the measured attenuation coefficients in the 20-keV image (Fig. 2) as a function of the KOH concentration in the holes.

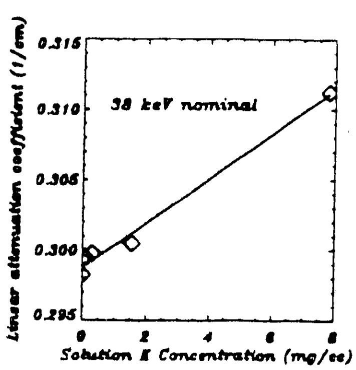

Fig. 5. As in Fig. 4, except for the 38-keV image (Fig. 3).

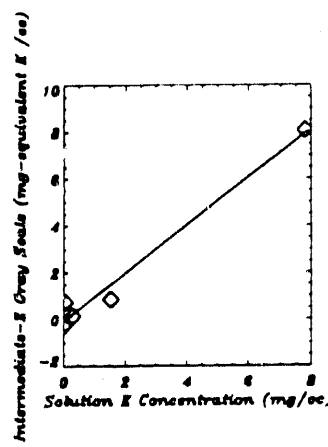

Fig. 8. Least-square fit of the measured intermediate-Z element group concentration in Fig. 6 as a function of KOH concentration in the phantom channels.

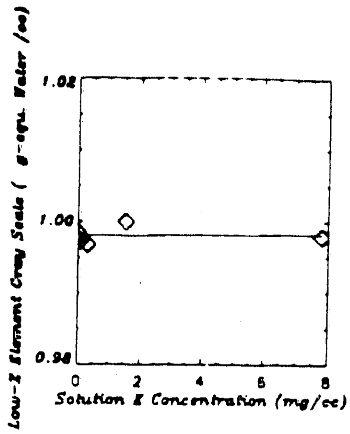

Fig. 9. The same as Figure 8, except for the low-Z element group concentration in Fig. 7.

We anticipate that small changes in local brain potassium concentration, such as that in focal hypercellularity of brain tissue caused by inflammatory cell infiltrates, will be detectable by MECT.

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