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Preliminary experience with monoenergetic photon mammography

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ABSTRACT

We are using a beam port at the National Synchrotron Light Source facility at Brookhaven National Laboratory as a source of monoenergetic photons. The photon source is radiation from a bending magnet on the X-ray storage ring and provides a usable X-ray spectrum from 5 keV to over 50 keV. A tunable crystal monochromotor is used for energy selection. The beam is 79mm wide and 0.5 mm high.

We imaged the ACR mammography phantom and a contrast-detail phantom using a phosphor plate as the imaging detector. Phantom images were obtained at 16, 18, 20 and 22 keV. Phantom thickness varied from 15 mm to 82 mm. These images were compared to images obtained with a conventional dedicated mammography unit.

Subjective preliminary results show that image contrast of the monoenergetic images is similar to those obtained from the conventional x-ray source with somewhat sharper and cleaner images from the monoenergetic source. Quantitative analysis shows that the monoenergetic images have improved contrast compared to the polyenergetic derived images. Entrance skin dose measurements show a factor of 5 to 10 times less radiation for the monoenergetic images with equivalent or better contrast. Although there remain a number of technical problems to be addressed and much more work to be done, we are encouraged to further explore the use of monoenergetic imaging.

1. INTRODUCTION

The conventional source of x-rays for medical imaging is the x-ray tube which generates a mixture of bremsstrahlung and characteristic x-rays. In modern mammography x-ray tubes, the target material usually used is molybdenum (Z=43). The characteristic peaks at 17 and 19 keV are reported to contribute about 25% of the photon flux, the remainder being a continuum of energies [1]. Other investigators [2] have experimented with different target/filter materials, Mo/Mo, Mo/Rh and Rh/Rh to improve the image contrast of the dense breast. However, there are trade offs in contrast and radiation dose as the effective energies are increased for better penetration of dense breast tissues.

Recently Boone and Seibert [1] did a computer simulation to compare performance of monoenergetic x-rays to polyenergetic x-rays from tungsten anode systems with regard to imaging. Their conclusion was that monoenergetic sources exhibited a 40 to 200 % improvement in tissue contrast when imaging the chest with different contrast targets. Admittedly, soft tissue contrast benefited the least. Burattini, et.al, [3,4] recently published their work using synchrotron radiation to image both breast phantoms and specimens. They conclude that the images obtained with monoenergetic x-rays have higher contrast, better resolution and similar, or less, radiation dose compared to the conventional polyenergetic x-ray images.

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We have embarked on a pilot project using a monoenergetic x-ray beam from the National Synchrotron Light Source (NSLS) at Brookhaven National Laboratory to explore the potential of monoenergetic photons for mammographic imaging. The following is a summary of our experience to date.

2. EXPERIMENTAL SET-UP

The National Synchrotron Light Source (NSLS) is an experimental facility consisting of two electron storage rings. The vacuum ultraviolet (VUV) ring stores electrons at 750 MeV to produce infrared, visible and UV light. The x-ray ring stores electrons at 2,500 MeV and extends into a higher energy x-ray region, up to 50 keV.

In conducting these initial studies, we borrowed whatever instrumentation was readily available. The physical arrangement, shown in figure 1, was that of a typical physics experiment and not a polished clinical facility. For our project, we had access to beam port X27C where the x-rays are generated at a main bending magnet on the electron storage ring. The x-ray energies emerging from the beam port range up to 50 keV. The beam is fan shaped with a width of 79 mm and about 0.5mm thickness.

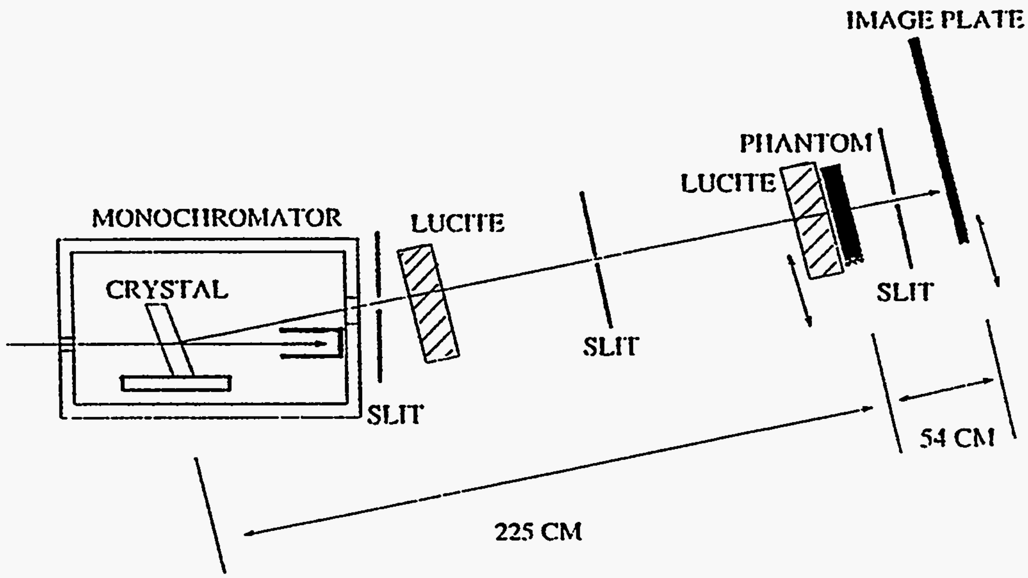

Figure 1. Experimental Setup

2.1 Monochromotor/beam scanning system

The monochromotor beams are produced by a Si(III) Laue monochromotor. This monochromotor produces a beam with a bandwidth of about 1.2 x 10-4 $\Delta E/E$ in the energy range used here (16 - 22 keV). The energy is set by the angle the crystal makes with the incident synchrotron beam (the Bragg angle). This angle is set remotely by computer to select the desired energy. There is a small contamination of this flux from the monochromotor due to harmonics of the fundamental energy. The largest harmonic contribution occurs at 3 times the fundamental energy and is ~ 0.1% of the fundamental intensity. After passing through the various absorbers used in the experiment, beam hardening effects will increase the relative intensity of the harmonic to the fundamental to only a few percent.

Collimation of the beam was placed at the exit of the monochromotor to shield against stray radiation, another slit was located approximately 60 cm in front of the phantom to be imaged, and a final slit between the phantom and the imaging plate serves as an anti-scatter slit. The overall length of the system from monochromotor to detector plate was 2.8 meters. To form an image, the imaging plate and the phantom were scanned through the beam. The total scan field was 79 mm by 87mm. This was large enough to cover about 2/3 of the phantom that we used. The drive system was a stepping motor and the translation speed for most of the images that we obtained was set at 2 mm per second.

2.2 Detector

The detector used to form the image was a Fuji ST3 PSP plate. This is a high sensitivity plate and the only one available for our use that could be read out with the model 2000 Fuji reader. A manual "latitude" control allowed 4 possible selections, from "1", which uses the 10 bits over a narrow window of the dynamic range to "4" which is a wide window and covers the entire range. All the phosphor plate readers installed at NSLS were optimized for x-ray crystallography work with high sensitivity and lower contrast settings than usually used in medical imaging. These plates were read out at a 2000 x 2000 matrix (100 μm /pixel) resolution. Since our interest was contrast resolution, we were not concerned at this point about the lack of spatial resolution, and we felt these plates would be adequate for our preliminary experiments.

2.3 Phantoms

We used two different phantoms for most of the experiments. The ACR phantom, model RMI-156, and a contrast-detail phantom obtained from the Sunnybrook Health Science Center, University of Toronto[5]. The ACR phantom was chosen since it is the standard for comparison of mammography and the accepted phantom for accreditation. The contrast-detail (CD) phantom allows for quantitative measurements of contrast as a function of spatial resolution. Figure 2 shows the detail of the CD phantom.

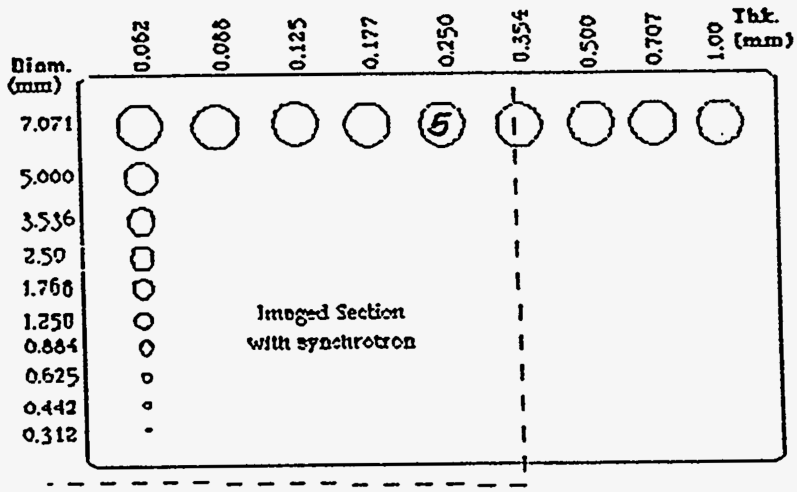

Figure 2. A schematic of the Contrast Detail Phantom used in these experiments.
Note that the portion imaged by the synchrotron beam is indicated by the dotted lines.
Lesion #5 is indicated

To assess the image contrast for different thicknesses of phantom, various layers of Lucite were placed in front of the phantom facing the x-ray beam. In the case of the monoenergetic beam and image plate detector, it was desirable to maintain a constant photon flux at the detector. This was accomplished by placing a total of 67 mm of Lucite absorbers in the beam plus the 15mm thick CD phantom. For a phantom image of 15mm, 67mm of Lucite were placed at the monochromotor, and the radiation entrance dose measured at the front surface of the CD phantom. To create thicker phantoms, a given thickness of absorber was moved from the monochromotor to in front of the CD phantom. Under these conditions, excluding variations in scatter, the photon flux should be constant at the imaging plate, the variation being the radiation entrance dose at the surface of the imaged phantom, and the increased scatter generated in the phantom as thickness increased. Figure 1 is a diagram of the experimental setup used with the synchrotron radiation beam.

3. EXPERIMENTAL PROTOCOL

3.1 Conventional x-ray mammography

For comparison to the conventional x-ray source, the CD phantom was placed on the Buckey platform above the film cassette. Additional Lucite thicknesses were added on top of the phantom facing the x-ray tube. To maintain approximately the same photon flux to the film, phototiming was used and the kVp and density control adjusted to provide an average film optical density of 1.2 for all images. Ion chamber measurements were made at the entrance surface of the phantom / absorber combination.

The contrast-detail phantom was imaged with a total phantom thickness of 15mm, 24mm, 44mm, 64mm, and 84mm. The ACR phantom was imaged with no additional Lucite thickness added.

3.2 Synchrotron Images

The beam size of the synchrotron beam was 79 mm wide at the phantom. The CD phantom is 150mm in width. Since we were interested in the most challenging contrast, we scanned the last 5 rows on the low contrast side of the phantom. Phantom thicknesses of 15mm, 24mm, 42mm, 62mm, and 82mm were used. The difference of 2mm between the monoenergetic study and the conventional study was due to the particular combination of Lucite absorbers available at NSLS. Each phantom thickness was imaged at 16 keV, 18.2 keV, 20 keV and 22 keV. The ACR phantom was imaged at each of the above energies with no additional thickness added.

TLD measurements were made at the entrance surface of the phantoms for each phantom thickness and for each energy. Ion chamber measurements were made in the beam as a function of absorber thickness and radiation dose at the surface of each phantom thickness was calculated from these measurements.

4. RESULTS

An example of the conventional x-ray image of the contrast-detail phantom at 44 mm thickness is shown in figure 3. A synchrotron image of the same phantom at 20 keV is shown in figure 4.
The largest and most visible target in these images is lesion number 5 as indicated on the phantom schematic. We purposely imaged the low contrast side of the phantom.

Quantitative measurement of contrast over lesion #5 (see diagram of the contrast-detail phantom) show that the contrast in the conventional image is on the order of 0.9%, measured from a digitized version of the film, compared to a contrast of 1.58% measured from the 18.2 keV and 1.12% from the 20 keV synchrotron images. All measurements were for the 42 mm thickness phantom.

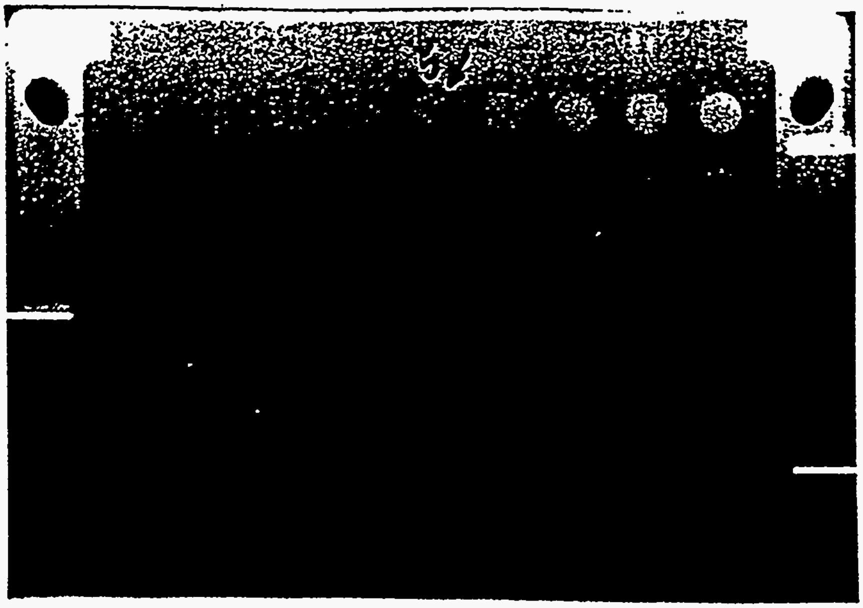

Figure 3. Image of the CD phantom 44 mm thickness at 28 kVp

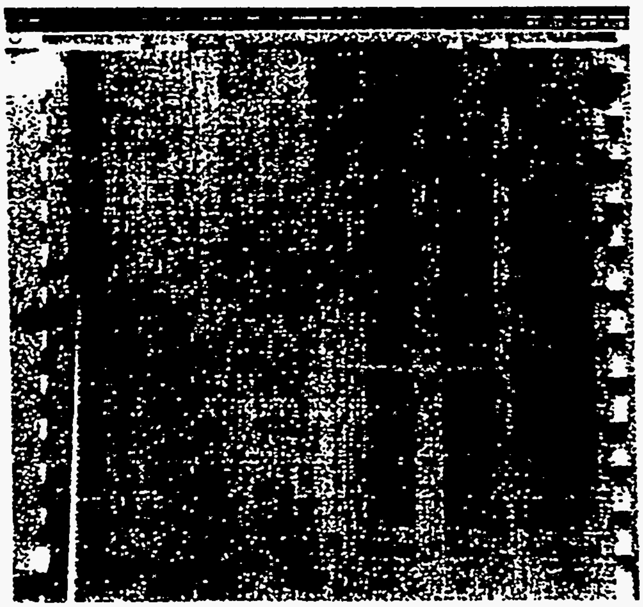

Figure 4. Image of the CD phantom 42 mm thickness with a synchrotron beam of 20 keV.

The radiation entrance dose at the surface of the phantom for various thicknesses and energies for the synchrotron images and the radiation dose for conventional mammography are shown in table 1. In the case of conventional mammography the mAs and kVp were varied to maintain an average optical density of 1.2 on the film. In the case of the synchrotron source, the photon flux at the imaging plate was held approximately constant by maintaining a constant total absorption in the beam path. A characteristic of the synchrotron is that the electron beam current decays with time which also changes the photon flux. The radiation doses reported in table 1 are normalized to a beam current of 75mA.

TABLE 1
Radiation Exposure to the Phantom Surface

Phantom Thickness(mm)	x-ray dose (mR)	Synchrotron radiation dose (mR)			
		16 keV	18.2 keV	20 keV	22 keV
15	78	-	6	-	-
24	158	-	10	-	-
42 (44)	758	108	93	121	38
62 (44)	2400	1080	241	-	-
82 (84)	4460	-	1245	1760	532

Thickness values in parentheses are for the conventional x-ray measurements

5. DISCUSSION:

The goal of these preliminary experiments was to determine if there was any advantage of using monoenergetic photons in imaging low contrast targets in mammography. Theoretically we should be able to improve the visibility of small differences in low Z targets. The properties of the monoenergetic photons, coupled with the beam characteristics that maintain a tightly collimated beam over many meters distance, should show contrast resolution close to the theoretical limit. The few other experiments that have been done confirm that such is the case [4,6].

A calculation of the contrast that should be obtainable for a 0.25mm thick Lucite target imbedded in 42mm of Lucite for different monoenergetic photons is shown in table 2. The contrast is calculated using the measured attenuation coefficients for Lucite at the energies used. Contrast is defined as the difference in attenuation through the target and surround divided by the attenuation through the surround. A calculation for a polychromatic beam of photons with 40% of the photons having 17 and 19 keV, and the remaining 60% of the photon flux spread in 2 keV energy bins from 14 to 28 keV, see figure 5, yields a contrast of 1.35%. The increase in contrast due to the use of monoenergetic beams with a line scanning detector system should yield images with improved contrast.

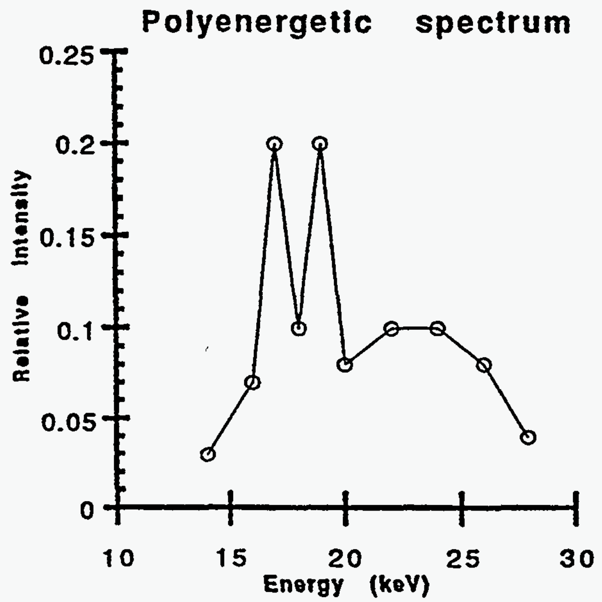

Figure 5. The simulated x-ray spectrum from a Mo target x-ray tube.

TABLE 2

Energy (keV)	16	18.2	20	22	poly-energetic
u cm^{-1}	1.019	0.716	0.648	0.536	--
Measured % contrast		1.58	1.12		0.9
Calculated % contrast	2.4	2.0	1.6	1.4	1.35

Note: % contrast is determined for a Lucite target of 0.25mm thickness in a Lucite phantom of 42mm thickness.
% contrast = $100(I_{41.75}/I_0 - I_{42}/I_0)/(I_{41.75}/I_0)$. Where I_x is the transmitted photon flux through x thickness of Lucite.

Our experiments are encouraging in that we could demonstrate an improvement in contrast in the phantoms. Our preliminary synchrotron images are similar to somewhat better in appearance to those that can be obtained with conventional polyenergetic photons and we expect to be able to improve the image quality with improved instrumentation. The major sources of problems are; 1) the monochromotor that we used introduced extraneous streaking and non uniformity into the image data. 2) the Fuji plate /readout system we used was not optimum for low contrast targets resulting in improper scaling of the 10 bits of available data. All the phantom-target data were confined to a small range of data out of the 1024 available.

We propose to carry out additional experiments with an improved monochromotor and different detector. Under consideration at this point is the use of mammographic film, or, preferable, a single line scanning digital detector. We are optimistic that with improved instrumentation, coupled with the advantages of the monoenergetic photon beam and the narrow beam geometry, we can show significantly improved contrast images.

6. ACKNOWLEDGMENTS

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