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The imaging results reported in the paper by Rubenstein [1] and elsewhere [2] were made possible by a careful choice of imaging methods as well as imaging equipment. As mentioned by Rubenstein, the quality of the images created was limited by the X-ray intensity available from the eight-pole 1.8 T wiggler magnet at the Stanford Synchrotron Radiation Laboratory (SSRL). [3] Hence, it was critical in the design of the imaging system to make the most efficient use possible of the available flux.

Since the heart is in constant rapid motion, data from each pixel of the image must be acquired at both of the required energies in less time than it would take the heart to move a fraction of the pixel width. Otherwise, significant blurring of the image would occur. However, for the best statistical accuracy in the imaging data, the greatest possible number of X-ray photons must be incident on each pixel. These two requirements of short exposure time and high photon count are obviously conflicting, and the only way to optimize both is to concentrate as much of the available X-ray flux into as small a part of the image as possible at any one time. Theoretically, the ideal way to do this would be to focus the entire intensity of the X-ray source into a pencil beam exactly the size of the desired pixel, and to raster this beam across the patient in a way similar to the electron beam in a TV tube. If the image has a size of 256 x 256 pixels as do our present images, such a pencil beam geometry could achieve the same experimental accuracy in the measurement of each pixel of the image as would a system with the X-ray flux spread out over the whole area of the image, but with an exposure time for each pixel 65536 times shorter. For a pencil beam geometry, therefore, blurring could easily be made entirely negligible, but the total exposure time for each line of the image could not be made too long or each successive line of the image might show the heart in a different position. Unfortunately, currently there is no way to produce such a pencil beam of synchrotron radiation at 33.16 keV or how to deflect it so rapidly. The next best technique is to focus the X-ray flux into a line, one pixel high and as wide as the full image. This is the natural geometry of a wiggler synchrotron source, and is easy to realize. For such a line-scanning arrangement, again with images of 256 x 256 pixels, the same experimental accuracy could be achieved as with an area-beam geometry with an exposure time per pixel 256 times shorter. Both the rastered-pencil-beam and line-scan imaging systems would have the same scanning speed requirement, that each line of the image be taken in a time short compared to the time needed for the heart to move the distance of the height of one pixel. In the raster case, such a scan speed would be needed only to insure that a continuous image of the heart is created, while in the line-scan case, the scan speed is needed also to prevent blurring.

The disadvantage of concentrating the X-ray flux into only a part of the image is that the whole image is no longer acquired at the same time. The result of this lack of simultaneity is a distortion rather than a blurring of the image. The line-scan case mentioned above is similar to the focal-plane shutter of a camera, for which a swinging golf club, for example, appears curved. For the specific case of coronary angiography, distortion is much less important than blurring. The distortion will only make an artery look more curved than it is, where blurring may make it look too wide, making arterial narrowing due to disease hard to discern.

As a result of these considerations, a line-scanning technique was chosen for the work at SSRL, and a means for achieving relative vertical motion of the patient and the wide horizontal X-ray beam had to be found. Because of the difficulty of moving the X-ray beam, which would require the rapid synchronized motion of the electron beam in the storage ring and of the monochromator crystals, it was decided to design a patient support chair which could both accurately position the patient at the correct angles with respect to the X-ray beam, and move the patient up and then down through the beam at a precisely known constant speed.[4] The patient chair uses a 2.5 HP stepper motor turning a recirculating-ball lead screw to move the patient vertically at up to 24 cm/sec. Part of the weight of the patient, as well as the weight of the chair itself, is counterbalanced with a gas-filled spring. The horizontal adjustment of the patient position perpendicular to the X-ray beam is also accomplished with a stepper-motor-driven lead screw. Both stepper motors are interfaced to the data acquisition and control computer, so that the patient can be accurately aligned based on the results of positioning images taken with the apparatus. In addition, two angular adjustments of the patient can be made with manually controlled DC motors.

As described by Brown,[5] the monochromator used for angiography at SSRL consists of two asymmetrically cut Su<1,1,1> crystals, which produce two line beams, one at an energy above the K-edge of iodine at 33.16 keV, and one with an energy below. As shown in Figure 1, the X-ray beams leave the monochromator displaced from each other by 3 mm vertically, and nearly or exactly overlap at the patient. This overlapping geometry is crucial to keep the time per pixel for the dual-energy exposure as short as possible. Since the patient must move through the beams, any displacement of the beams relative to each other will lead to a delay between the two energy exposures and to possible misregistration artifacts.

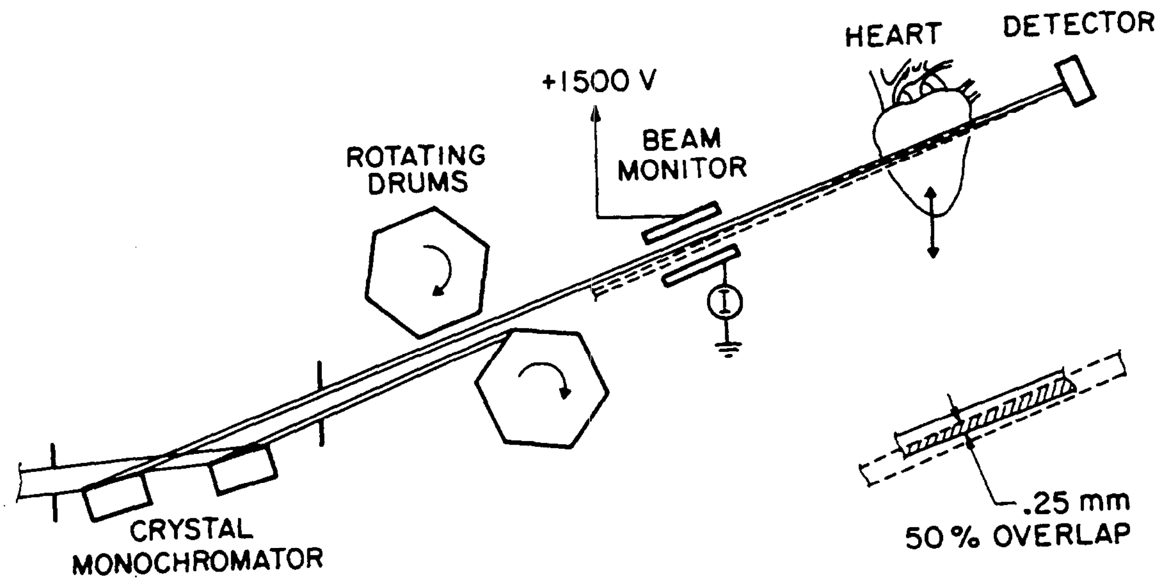

Figure 1a. The experimental arrangement for synchrotron radiation angiography used at SSRL for human subject studies. Dual asymmetrically cut silicon monochromator crystals are used to produce separate fan-shaped X-ray beams, one with an energy above and one with an energy below the K-edge of iodine at 33.16 keV. Rotating ten-faceted aluminum drums are used to alternatively occlude one beam and then the other. A single argon filled ionization chamber is used to monitor the incident X-ray flux. A single 300 channel SI(Li) detector measures the transmitted flux.

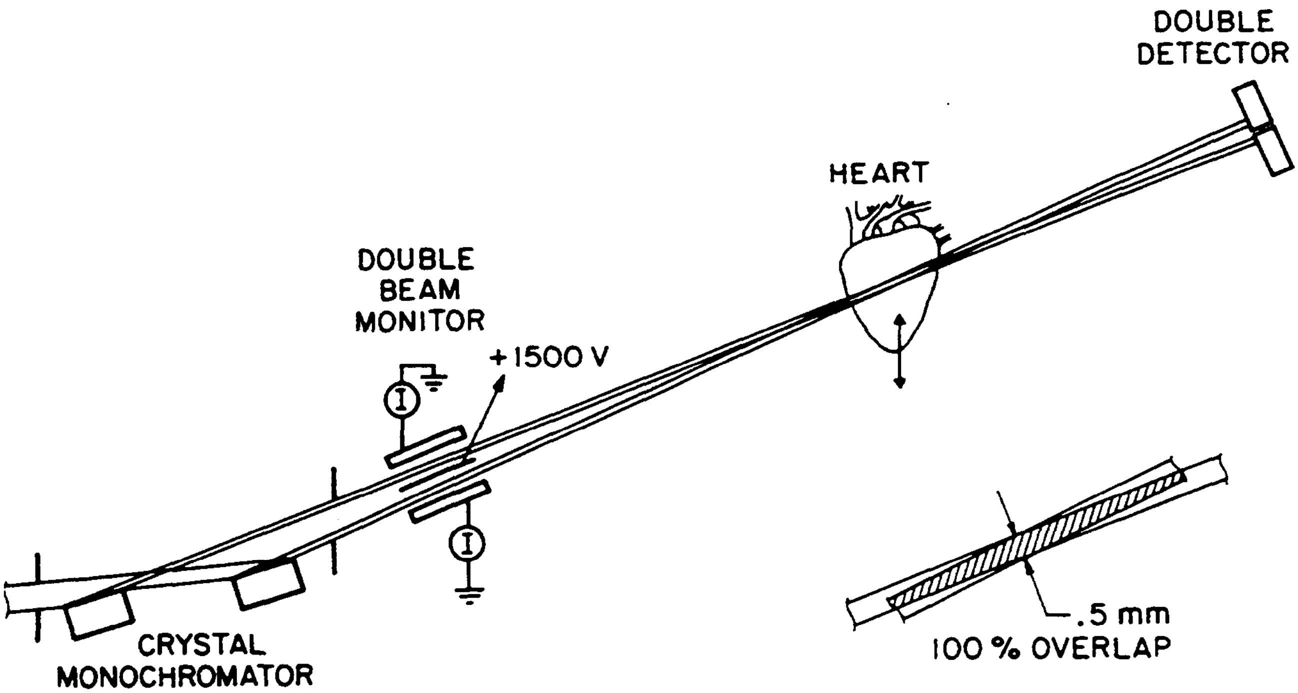

Figure 1b. The experimental arrangement for synchrotron radiation angiography for future use at SSRL and at NSLS for human subject studies. The separate X-ray beams are monitored with a double ionization chamber, and pass simultaneously through the patient. The transmitted flux is measured with a Si(Li) detector with a dual array of 300 channels.

Two different imaging techniques are shown in Fig.1. The method shown in Figure 1a has been used up to now at SSRL, since only a single array Si(Li) detector was available to measure the transmitted flux. In this setup, two ten-faceted tin-plated aluminum drums are rotated in synchronization with each other and with the patient chair motion in such a way that first one beam and then the other is occluded by the drums.[6] Each drum is belt-driven by a 5 HP AC induction motor. The speed of each motor is controlled by a variable frequency motor driver. The two motor drivers are controlled by a DEC LSI 11/02 computer by means of a custom built interface. This software-controlled phase-locked-loop system is able to control the synchronization of the rotating drums to about 0.2 msec. The technique shown in Figure 1b, in which a double beam-monitor and a dual-array Si(Li) detector is used, was tried with a prototype 3 cm wide, 120 element detector in January, 1987. A 600 element dual-array Si(Li) detector will be used for human patient studies for the first time early next year.

Because the intensity of the synchrotron beam is not exactly constant, accurate beam intensity monitoring is crucial to achieving the greatest possible image quality. Argon-filled gas ionization-chambers have been built as beam monitors for both the techniques shown in Figure 1. The high voltage electrodes in both chambers are mounted on large plastic insulators so that voltages as high as 3 kV can be used to give fast response. For a truly monochromatic X-ray beam, such a beam monitor should be able to exactly correct for any beam intensity fluctuations. However, to the extent that off-energy contaminants are present, an unavoidable error in the beam monitoring will occur. This is due to the fact that a relatively thin ionization chamber will produce a larger signal from a lower energy photon than from a higher energy photon, due to the larger interaction cross-section at lower energies, while a totally absorbing detector will produce a larger signal from a higher energy photon than from a lower energy photon due to the greater absorbed energy from the higher energy photon. Because off-energy contaminants tend not to be as well collimated as the monochromatic beam, placing the beam monitor as close to the patient as possible reduces this source of inaccuracy.

The overall accuracy of the imaging system is in a large part dependent on the noise performance of the X-ray detector used. Consequently there is a justification in spending considerable effort in designing and building the best possible detector. At SSRL we have used a multi-channel lithium-drifted silicon detector [7] with 300 detector electrodes mounted 0.5 mm apart on a single 5 mm thick piece of silicon. The detector is mounted in a vacuum chamber and is cooled to -20° with a thermoelectric cooler. A potential of 500 V is applied to the single positive electrode, while each of the 300 negative electrodes is connected to ground through its own channel of signal conditioning electronics. This detector system has the advantage of being a very simple physical system, with almost ideal properties of very low veiling glare [8] and very fast response time. In the tests in January, 1987, the detector was measured to have an accuracy determined entirely by the statistical noise of the absorbed X-ray photons over the entire dynamic range from the unobstructed primary X-ray beam to a beam passing through the equivalent of a thick human patient. Considering that the quantum efficiency of the detector is about 70%, this performance is about as near to the theoretically ideal performance as is possible. The silicon detector, although a very simple physical system itself, requires a large, expensive, and complex read-out electronic system.[9] In order not to degrade the performance of the detector, the read-out system must be very stable, highly linear, and have sufficient resolution. These criteria were met in this system by using a gain switching amplifier in combination with a high accuracy 5 MHz voltage to frequency converter for each detector channel.

The present system uses two integrated circuit operational amplifiers and three relays per channel as a current to voltage converter with variable gain in binary steps over a range of 1 to 128. A digital to analog converter is used for each channel to adjust the pedestal.

Both the relays and the DAC's are remotely controlled by the data acquisition computer. The voltage signal from the second operational amplifier is integrated using a voltage to frequency converter, and the resulting pulse train is counted using LeCroy Model 4434 32 channel CAMAC scalers. At present we have 256 channels of electronics connected to a 300 channel silicon detector. Since the electronics is mounted in single width NIM modules at 4 channels per module, these 256 channels including power supplies require two full size electronics racks.

For the dual detector with two rows of 300 channels, which is presently under construction at the Lawrence Berkeley Laboratory (LBL), and for the 0.25 mm resolution detector which is planned to have two rows of 600 channels, a more compact and lower cost form of the data acquisition electronics has been designed at LBL. This system will have 10 channels per circuit board, and will have scalers and DMA data-acquisition circuitry included, obviating the need for a CAMAC system. The full 1200 channel system should fit into the same two racks that presently house a 256 channel system.

Data must be acquired from the detector and the beam monitor in the correct manner to allow for corrections for variations in intensity across the X-ray beam profile, for variations in sensitivity between different detector and electronics channels, for drifts in the pedestals for each detector channel, and for noise pickup from the 60 Hz line. To eliminate the effects of 60 Hz pickup, all data is acquired in synchronization with the 60 Hz line. In the present system, this synchronization with the line is accomplished in three ways. The rotation of the X-ray beam chopping drums is synchronized with the line by means of the digital phase-locked loop. The data acquisition is synchronized with the rotating drum motion by means of trigger pulses generated by the phase-locked loop control interface. The vertical scan speed of the patient chair is accurately set to correspond to the scanning rate determined by the rotating drums. Pedestal measurements are made before and after each scan of the patient through the X-ray beam, for each different phase of the 60 Hz line, by closing a mechanical X-ray beam shutter. Before and after the patient enters the X-ray exposure hutch, calibration data are taken, with the shutter closed, with the X-ray beam unobstructed, as well as with the beam passing through a variety of different aluminum absorbers. All calibration data are taken for each phase of the 60 Hz line as well as for each facet of each rotating drum. After the patient imaging data is acquired, an image generation program combines the imaging data and calibration data to create accurate digital images for each frame of the patient study.

For the data acquisition, experiment control, image generation, image processing, and image archiving functions, a computer with adequate speed, a large magnetic disk drive, and a high speed magnetic tape drive is essential. Although for an operating angiography clinic these functions would ideally be divided up between a number of different machines, each dedicated to a specific task, a single machine has been used for the SSRL project, with good results. The computer used with this experiment is a DEC LSI 11/73 with a Fujitsu Eagle 450 Mbyte disk drive and a Cipher 3200 BPI magnetic tape drive. For ease in real-time programming, the RT-11 (version 5.01) operating system is used. For the 0.5 mm resolution images used up to now, this computer is totally adequate for data acquisition, image generation, and image archiving functions. For image processing, however, a more powerful computer is desirable. A great deal of speed improvement has been achieved by writing all image processing programs in PDP-11 assembly language, but the most favorable algorithms still require about 15 minutes per image and use the full address space available to the PDP-11 processor. For 0.25 mm resolution images, which will require four times as much data, and take eight times as long to process, a computer both with faster processing speed and with the ability to directly address a larger RAM memory would be essential.

A radiation safety system has been built to prevent exposure of the patient or physician to unsafe conditions. It is based on a pair of fast closing shutters which are controlled by a radiation interlock control system. Two totally independent chains of interlocks are used and a fault in either chain rapidly turns the beam off.

Some of the unsafe conditions which create a fault are:

1. At SSRL the rate of X-ray exposure from the monochromatized beams is about 50 R/s. Thus, the shutters must never be open if the patient chair is not moving at the scan speed.

2. Since an upper limit to beam intensity is known from SSRL operating conditions, total patient radiation exposure can be kept within safe limits by limiting the total length of a scan sequence (typically 25 s).

3. The physician initiates a scan by closing and holding a SCAN switch. If he observes some malfunction in the scan operation, release of the SCAN switch immediately extinguishes the beam.

Each safety shutter consists of a pair of rotating blades held open by two rotary solenoids. When the solenoids are de-energized, the blades close within 50 ms. Each safety shutter has an effective thickness when closed of 2.30 mm of lead, 2.30 mm of steel, and 1.15 mm of stainless steel.

The examination room is also protected from any penetrating gamma radiation from the primary beam direction and only monochromatized X-ray beams of 33 keV +/- 1 keV can enter the examination room.

To measure the total radiation exposure received by the patient, a calibrated radiation monitoring system has also been implemented.

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