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A 1200 Element Detector System for Synchrotron-Based Coronary Angiography

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Abstract

A 1200 channel Si(Li) detector system has been developed for transvenous coronary angiography experiments using synchrotron radiation. It is part of the synchrotron medical imaging facility at the National Synchrotron Light Source. The detector is made from a single crystal of lithium-drifted silicon with an active area 150 mm long x 11 mm high x 5 mm thick. The elements

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are arranged in two parallel rows of 600 elements with a center-to-center spacing of 0.25 mm. All 1200 elements are read out simultaneously every 4 ms. A Intel 80486 based computer with a high speed digital signal processing interface is used to control the beamline hardware and to acquire a series of images. The signal-to-noise, linearity and resolution of the system have been measured. Human images have been taken with this system.

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1. Introduction

Progress in heart research is limited by the danger to the patient of the current methods of imaging coronary arteries. The only currently available method makes use of an arterial catheter which is inserted into a peripheral artery and then advanced up to the heart. A fluoroscope is then used to position the catheter at the entrance to either the left or right coronary artery. A contrast agent containing iodine is then injected through the catheter into the artery and a series of x-ray images are taken. Since most of the contrast agent goes into the coronary artery, excellent quality images are produced. Unfortunately, the use of an arterial catheter makes this a dangerous procedure so that it is only done when absolutely necessary. Use of the special characteristics of x-rays from synchrotron radiation sources may permit the development of much safer imaging techniques than would be possible with traditional x-ray sources.

A much safer alternative to arterial catheterization is the injection of the contrast agent into a vein. The disadvantage of this technique is that the contrast agent must flow through the heart, out the pulmonary arteries to the lungs, and back through the pulmonary veins into the other side of the heart before it reaches the coronary arteries. This results in the contrast agent being diluted by a factor of at least 20:1 by the time it reaches the coronary arteries. Standard angiography instrumentation is unable to produce an adequate image at this level of contrast dilution. However, synchrotron radiation allows one to make use of details of the iodine x-ray absorption spectrum rather than just the total absorption. In particular, it is possible to produce a pair of x-ray beams with energies on either side of the iodine K absorption edge and take a simultaneous pair of images with each beam. Since the absorption coefficient of iodine increases by a factor of about 6.5 at the K edge while all other elements are almost unchanged, the only substantial difference in the two images will be due to the iodine contrast agent. Thus, a subtraction of one image from another will result in a difference image displaying the location of the contrast agent with minimal sensitivity to intravening tissue and bone. The high intensity of the x-rays within the energy bandwidth

of the two beams allows the difference image to have an acceptable signal to noise ratio.

2. Experimental Setup

An experimental facility for medical imaging has been developed at the National Synchrotron Light Source (NSLS) on the X-17B2 beamline by a team from Stanford University, NSLS, Lawrence Berkeley Laboratory, North Shore University Hospital, Palo Alto Veterans Hospital, and Stanford Synchrotron Radiation Laboratory. [1-3] The layout of the system is shown in Figure 1. A superconducting wiggler magnet (currently run at 4.0 Tesla for angiography) on the NSLS produces a horizontal fan of white radiation. This radiation is incident on a bent Laue monochromator [4-6] which with the use of slits and beam stops produces a pair of horizontal x-ray fan beams 125 mm wide that cross at the patient at a distance of 37.4 m from the center of the wiggler. The patient sits in a motorized chair that is scanned vertically in front of the beam at a speed of 120 mm/s which when combined with a detector integration time of 4.17 ms per image line results in a vertical pixel size of 0.5 mm. This is a good match to the vertical size (0.4 mm) of the beam. For 250 horizontal image lines, this results in a total of 1 second of imaging time. 1.2 seconds is consumed in decelerating and reversing the motion of the chair, so the start time of adjacent image frames are 2.2 seconds apart. The chair has additional motors to change the horizontal position of the patient, and to perform roll and yaw rotations on the patient. The detector is located 3.7 m behind the patient. At this point, the beams are vertically separated center to center by 3.5 mm. Each beam has an energy bandpass of about 135 eV and the centers of the bandpasses are separated by about 300 eV.

Preparation for a set of patient images begins before the arrival of the patient at the experimental facility. Initially, the correct alignment and intensity of the x-ray beams is verified. Then, test images are taken using a phantom containing a pig heart with a bone column to check the visual appearance of images. Once this step is complete, a series of calibration scans is taken using a uniform water phantom in a plastic tank. These calibration

images are used to normalize the final patient images to take out the effect of variation in the response of channels across the horizontal spread of the beam. The calibration images are taken at several different settings of detector electronics amplifier gain in order to cover the range of amplifier gain settings expected to be used for the patient.

Once the patient arrives, a venous catheter is inserted into the patient in a room adjacent to the main imaging room, using a fluoroscope to show the position of the catheter while it is being positioned. When the physicians are satisfied with the location of the catheter, the patient walks to the imaging chair and a power injector is connected to the catheter. Then, the physicians use manual controls on the chair to rotate the patient to an approximate starting orientation. At this point, a positioning scan is taken with the x-ray dose reduced to 10% of the normal dose using only a small amount of contrast agent (usually 5 ml) and with only one image frame. Image frames are automatically copied from the data acquisition computer to the image processing workstation. The positioning scan is examined on the workstation by the physicians to determine any adjustments to the patient scan path required to better position the heart in the image frame. This image is also used to estimate the range of image pixel values that will be in the final patient images. A special program on the image processing workstation aids in determining the amount that the detector amplifier gain should be changed by to make the best use of the dynamic range of the detector. At this point, the final patient scan is taken. For the final patient scan, an x-ray exposure of around 5 rem per frame is given and a larger bolus of contrast agent is injected (typically 35 ml at 15 ml/s). This bolus takes about 8 s to reach the coronary arteries at which point imaging is generally started. Typically, about 2 to 4 image frames are taken. The best images are usually around 9-11 s after the injection of contrast agent. This sequence of images is analyzed by the physicians on the workstation and the next orientation for patient imaging is decided. After a wait of about 30 minutes for the contrast agent to clear out of the patient's bloodstream, the sequence of positioning scan and imaging scan is repeated with a different orientation of the patient.

3. Detector

The detector itself is made of a single, large crystal of lithium-drifted silicon ($\text{Si}(\text{Li})$) and is the latest in a series of detectors built for this project. [7] A schematic of the detector is shown in Figure 2. It has 1200 detector elements with a center-to-center spacing of 0.25 mm which determines the horizontal pixel size of the detector. The vertical pixel size is 0.5 mm. The 1200 elements are divided into two parallel rows of 600 elements each which are separated by 0.2 mm. The active area of the detector is 150 mm wide by 11 mm high by 5 mm thick. The signals from the detector elements are sent by a set of twisted pair cables into an adjacent room where the detector electronics is located. All 1200 elements of the detector are measured every 4.17 ms.

The electronics of this system measures integrated current rather than individual photons. Since the incident x-ray beam is essentially monochromatic, the integrated current is proportional to the x-ray flux. The first stage of the electronics for each channel consists of a two-stage operational amplifier with a programmable gain that can be changed by factors of two in 8 steps achieving a total range of 128. Each channel also has a programmable 8-bit digital-to-analog converter (DAC) that is added together with the input of the operational amplifiers to compensate for variations in the leakage current of each channel. Finally, a single 16-bit DAC is fed into the inputs of each channel for use in linearity measurements.

The output of the amplifier is fed to a voltage-to-frequency converter (VFC) and a pair of digital scalars. The VFCs are individually selected from production runs to have a linearity of 0.01% or better. The VFC system uses a reciprocal counting technique [quote reference] to achieve a dynamic range of 40,000:1 with a cycle time of 4 ms or less. The range of sensitivity of the electronics at 33 keV is from 163 photons per digital output count at the lowest gain to 1.3 photons per digital count at the highest gain. This corresponds to a current sensitivity of 0.60 pA per digital count at the highest gain. The digital output counts of each channel are buffered so that the next line of data can be acquiring while the previous line of data is being read out.

There is a small component of 60 Hz noise in the readout signal. For this reason, the data is acquired in synchronization with the 60 Hz power line voltage. Each cycle of the voltage is divided into 4 sections or phases giving a repetition rate for the data acquisition of 240 Hz and a corresponding measurement time of 4.17 ms. The average noise level for each phase is measured just before and after a series of image frames is taken. This noise level is averaged separately for each phase and is subtracted separately from data taken at the corresponding phase during the image processing after an imaging scan.

4. Data Acquisition and Analysis

The NSLS angiography beamline uses two primary computers for data acquisition and analysis. The data acquisition is controlled by an Intel 80486 based computer running MSDOS, while most of the data analysis is performed by a Unix workstation (currently a Decstation 3100 and soon to be an IBM RS/6000). This division of effort is motivated by the observation that the angiography data acquisition computer must have excellent real time response since the data must be acquired on a fixed time schedule in order to synchronize with the motion of the chair that the patient sits in. However, for angiography data analysis, it is desirable to use a fast computer with a versatile image processing environment. An Intel 80486 based MSDOS system was chosen for the data acquisition task, since the hardware is fairly inexpensive and because it is relatively straightforward to write programs that take complete control of the underlying hardware.

The data acquisition computer is interfaced to several different pieces of equipment as shown in Figure 3. These systems include a custom-built interface to the detector electronics for the Si(Li) detector. The computer uses a CAMAC to RS-232 interface to send commands to the stepping motors of the chair and to other stepping motors that move the Si(Li) detector and upstream slits. The position of the chair is measured using encoders that are read out with CAMAC up-down counters. In addition, a CAMAC digital I/O register opens and closes the patient safety shutters and reads the status of the patient safety system.

These systems will now be described in more detail.

By far, the most complicated interface is that to the Si(Li) detector electronics. The electrical interface is handled using a digital signal processing (DSP) board in the PC. It was made by Spectrum Signal Processing, Inc. and it uses a Western Electric DSP32C chip. This board is interfaced to an electronics interface box which then fans out command and data cables to the five large size VME crates. The DSP card interfaces to the PC using dual-ported memory that looks like regular memory to the DSP CPU, but like I/O ports to the PC CPU. The DSP32C CPU runs a program that is downloaded from the PC at the time the PC data acquisition program is started. The DSP normally spends most of its time monitoring a memory location in the dual-ported memory for the presence of a command character. The PC sends the DSP a command by writing a character to that memory location. The particular character written determines what command the DSP executes.

The most important DSP command is the 'u' command which tells the DSP to acquire a line of image data. The DSP does this by sending a command signal to the electronics interface box which tells it to latch a copy of the data from each channel of the 1200 channel detector into a matching buffer. The original data values are now zeroed out so that the detector electronics can go back to counting new data values. The DSP now reads out the latched copies of the channels by sequencing through the 1200 channel addresses in order to read the values. The DSP interleaves reading data values with converting them into detector count values, which are written into a 1200 element buffer in the dual-ported memory it shares with the PC. Finally, the DSP writes a null byte over the 'u' byte in the command character buffer to let the PC know that the DSP has finished the command. Other DSP commands exist to do things like read the AC power phase clock, synchronize with the zero crossing of the AC phase, and other miscellaneous things.

The chair that the patient sits in is moved by a stepping motor controller that is addressed by the PC through a CAMAC controlled RS-232 port. The position of the chair is measured by a pair of position encoders and feed into a pair of up/down CAMAC counters. An

independent safety system interlock verifies that the chair is moving at the correct speed. A second motor controller is used to control the vertical and horizontal position of the detector and a pair of beamline slits. In addition, the PC can open and close the photon shutter for the imaging room, trigger the contrast agent power injector, and query the status of the safety system.

The software for the data acquisition system was written using the Watcom C/386 protected mode compiler with a DOS extender to allow easy manipulation of the large arrays used for imaging. The primary function of the control program is, of course, to take X-ray images of a patient or phantom. To orchestrate this, the PC has to synchronize the operation of several different systems during a scan. Besides reading out data from the Si(Li) detector, the PC must ensure that the patient chair is at the correct position for the particular scan line being taken, that the X-ray shutters are opened and closed on the right scan lines, and that the patient safety systems have not shown a fault. It also records in the data file several additional values for each line including the chair position, the DSP phase clock value, the safety system status, the ion chamber monitor current, and the patient electrocardiogram signal. In addition, it takes pedestal measurements before the first image and after the last image. A special version of the PC control program is used for actual human patient imaging, which has the minimum functionality needed to take and display the images. This is in order to limit the amount of software that must be specially tested for use with a human patient.

The other functions of the PC control programs divide into two categories: manipulation of previously taken images and testing of various components of the system. Since the primary image processing is done on a Unix workstation, the PC's image manipulation features are limited to operations like, for example, contrast adjustment and use of the current image for positioning the next scan. The PC programs have a much wider variety of functions for testing the operation of system components. The most commonly used function continuously displays the count rate in each channel as a series of graphical bar graphs, one for each channel, with the bar graph display divided into one half for the high detector and

one half for the low detector. This may be displayed for all phases or for individual phases alone. A display of channel standard deviations is also available. Other functions include moving the patient chair and testing its operation, scanning the detector position relative to the X-ray beam to optimize its position, and optimization of the pedestals for each channel of the detector.

5. System Performance

A. Noise

There are several different kinds of phenomena that affect detector images that fall under the general classification of noise. Statistical fluctuation of the photon count is a fundamental limit to the counting resolution, but many of the other sources of noise are not fundamental and, in principle, can be corrected for. Such noise sources can include variation in the trajectory and intensity of the photon beam from the wiggler magnet and mechanical vibration of components of the beamline or the detector system. The most important component that can be dealt with is electrical noise in the system. The elimination of sources of electrical noise consumed a large part of the checkout phase of this detector system.

A detector channel with a relatively large amount of electrical noise is commonly called a "bad channel". If a bad channel is identified, steps are taken either to fix the channel electrically, or, if that is not practical, remove the effect of the bad channel in the image by replacing that channel with a value interpolated from channels adjacent to it in the image. The most common way to find bad channels is by taking a large number of image lines with no incident x-rays as a sample set. For each detector channel, the standard deviation of the channel count rate is computed. Then, the mean value of these standard deviations is computed and also the global standard deviation of the individual standard deviations. Any channel whose individual standard deviation exceeds the mean of the standard deviations

by some multiple of the global standard deviation (typically around 2 or 3) is regarded as a bad channel. Figures 4 and 5 show some example graphs.

A different kind of bad channel is one that has a dark current that is significantly different from the other channels. These channels show up during the process of pedestal optimization. Each detector channel has a programmable offset which may be used to shift the channel's output. The pedestal optimization process repeatedly reads all the channels used and tries to force them to have about the same value by adjusting the offsets. This is done to make the best use of the dynamic range of the detector. If the optimization leads to a channel's offset reaching the limit of its range at 0 or 255, the channel is probably bad, since, for this case, it is usually not possible for pedestal subtraction to produce a good looking image line.

The handling of bad channels in images where the detector is exposed to x-rays is complicated by the fact that the definition of a bad channel is count rate dependent. A channel that is only slightly bad may have no impact on the image if it is located in a section of an image with a high count rate. But in an area of an image with a low count rate, it may have a noticeable effect on the image. Human patient images cover a large dynamic range, so it is not simple to construct an algorithm that unambiguously identifies bad channels in a real image. For this reason, human patient images are routinely scanned by an operator using image processing software on the Unix workstation to identify bad channels in the image and remove them after the images have been acquired. With the current detector system, there are about 20 to 30 bad channels in an image.

B. Linearity

The linearity of this detector system is important, since the images showing the contrast agent are obtained by taking the difference of two sets of large numbers. Any significant nonlinearities will show up as artifacts in the difference image data taken from two different elements. The linearity of the various channels is measured by a programmable 16 bit calibration digital-to-analog converter (DAC) that is fanned out to all 1200 channels of the

detector system. Each channel of the detector adds this calibration DAC voltage to the voltage signal from the detector before sending on to the amplifier and voltage-to-frequency stages of the electronic system. This DAC is set to zero for normal x-ray imaging. A linearity test is performed by ramping up the calibration DAC in uniform voltage steps and fitting the data for each channel to a line. The result of the fitting shows that the channels have an average non-linearity of approximately 0.02 %, if bad channels are removed from the average.

C. Resolution

The resolution of detector images was tested with a fluoroscopy test target made by Nuclear Associates. The target contains four adjacent copper mesh strips with varying mesh sizes of 1.3, 0.75, 0.65, and 0.50 mm per wire surrounded by a helix of lead numerals and with all of this embedded in plastic. Figure 6 shows a section of an image taken of the test target. This section is centered on the part of the target containing the wire meshes. Since a pixel in the raw data corresponds to an area of 0.25 mm horizontally and 0.5 mm vertically, the aspect ratio of the image has been corrected to look like the actual object by stretching the pixel data vertically by a factor of two.

In Figure 6, wires running in both directions can be seen for the 1.3 and 0.75 mm per wire meshes. However, for the 0.65 and 0.50 mm per wire meshes, only features due to vertical wires can be seen. This reflects the fact that the detector intrinsically has more resolution in the horizontal direction than in the vertical direction.

D. Modulation Transfer Function

The modulation transfer function (MTF) of the detector was determined by two methods. The first method of measurement used a standard 0.1 mm Pb square-wave chart. The second method made use of the edge spread function as described in Dainty and Shaw [8]. using iodine to form the edge. The iodine was in the form of an 185 mg/ml iodine solution located

in a wedge-shaped cavity in a larger piece of Lucite. The measurements revealed that the MTF is not depending on the iodine signal contrast or the detector gain setting. Figure 7 shows a summary of the MTF as measured by the two methods.

E. Human Patient Images

Figure 8 is an image taken with the new detector system on June 3, 1993 at NSLS. This image is from an individual who has undergone directional atherectomy of a high grade proximal right coronary stenosis. The view is the equivalent of a 50deg left anterior oblique projection in a clinical angiogram. The aorta and left ventricle are mostly opacified. The left main coronary artery is also visible. The right coronary artery is visible throughout its entire length, including its terminal branches.

6. Conclusions

The 1200 channel detector is now installed in the NSLS angiography facility. Except for some bad connections in the detector cryostat, the detector has performed very well. In the near future, the problems with these channels should be corrected by changing a defective connecting circuit card. The noise of the detector system has been measured to be very close to the limits set by the counting statistics of the incoming beam once a correction for the fluctuation in the incident beam has been made. The measured non-linearity of the system is typically less than 0.02%. The modulation transfer function of the system has also been measured and it agrees with calculations based on the detector properties.

The quality of human images taken with this synchrotron-based system, while not yet as good as those taken with the invasive arterial technique, show that coronary arteries can be clearly seen with this less invasive, venous injection method. The NSLS angiography imaging facility is now ready to be used by a team of physicians to study a group of patients. This study is scheduled to start in the Fall of 1993 and a medical evaluation of the medical utility of this technique should be complete in late 1994 or early 1995. The significantly

reduced risk of this technique may permit studies of the progression of heart disease and might some day be useful for clinical studies.

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FIGURES

FIG. 1. Schematic diagram of synchrotron-based imaging system.

FIG. 2. Schematic of detector geometry.

FIG. 3. Organization of the data acquisition system.

FIG. 4. Fractional noise per channel of the high energy detector. The three curves show the original detector channel noise, the noise after normalization by the ion chambers, and the theoretical noise.

FIG. 5. Fractional noise per channel of the low energy detector.

FIG. 6. Image of a section of the fluoroscopy test target.

FIG. 7. Measured modulation transfer function for the 1200 element detector.

FIG. 8. Image of a patient taken with the 1200 element detector system.

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