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MASTER

A.C. Thompson, F.S. Goulding, H.A. Sommer, J.T. Walton, E.B. Hughes, J. Rolfe, and H.D. Zeman

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A. C. Thompson, F.S. Goulding, H. A. Sommer and J. T. Walton
Lawrence Berkeley Laboratory
University of California
Berkeley, California 94720 U.S.A.

and

E. B. Hughes, J. Rolfe and H. D. Zeman
Hansen Laboratories of Physics
Stanford University
Stanford, California 94305 U.S.A.

Abstract

A 30-element Si(Li) detector has been fabricated to measure the one-dimensional flux profile of 33 KeV x-rays from a synchrotron radiation beam. The device, which is fabricated from a single 39mm x 15mm silicon wafer, is a linear array of 0.9mm x 7mm elements with a 1mm center-to-center spacing. It is 5mm thick and when operated at room temperature has an average leakage current of 10 nA/element. The x-ray flux in each element is determined by measuring the current with a high quality operational amplifier followed by a current digitizer.

This detector is being used to study the use of synchrotron radiation for non-invasive imaging of coronary arteries. The experiment uses the difference in the transmitted flux of a monochromatized x-ray beam above and below the iodine K-edge. Measurements have been made on plastic phantoms and on excised animal hearts with iodinated arteries. The images obtained indicate that a 256-element device with similar properties, but with 0.6mm element spacing, will make a very effective detector for high-speed medical imaging.

Introduction

Synchrotron radiation x-ray sources provide intense, tunable beams which make possible many exciting areas of research which are not possible with conventional x-ray sources. These x-ray beams place very demanding requirements on detector systems since they are very intense, highly collimated and have a pulsed time structure.

We have developed a 30-element Si(Li) detector for high-speed flux measurements with good spatial resolution and high efficiency. This detector is a prototype for a larger detector which will be used for digital subtraction angiography using synchrotron radiation.

Angiography, the radiologic visualization of blood vessels and the chambers of the heart containing contrast media, has become a powerful method of examining the internal condition of the cardiovascular system and of diagnosing disorders involving the circulation of the heart, the brain, and of other vascular beds. However, the techniques currently in widespread use involve invasive procedures that carry significant risks of morbidity and mortality. For example, the diagnostic procedure for coronary artery disease requires the insertion of a catheter into the ostia of each of these arteries in turn and the injection of concentrated solutions of iodine-containing compounds. The risks of this procedure are far too high to permit the routine use of angiography to detect the presence of dangerous obstructing lesions in the coronary arteries in the many persons known to be at risk. Angiographic studies are limited to symptomatic patients and are not conducted on asymptomatic patients even if they are at high risk because of hypercholesterolemia, hypertension, family history, etc. There is, therefore, an urgent need to reduce both the risk and cost of these procedures.

Conventional x-ray sources provide a broad spectrum of x-ray energies. However, iodine preferentially absorbs at energies in the immediate vicinity of its K-absorption edge at 33.16KeV. In addition, other body structures, such as soft tissue and bone are reproduced in the image and detract from the contrast due to the iodine itself. If the x-rays from a conventional source are filtered to select the energies specific to iodine and thereby enhance the iodine contrast, there is insufficient intensity in the beam to produce a useful image. This is the reason, in essence, for the dangerous invasive procedure now in use.

The enormous intensity and inherent collimation of synchrotron radiation permit it to be energy monochromatized by Bragg diffraction to produce a fan beam of excellent energy resolution ($\Delta E/E \approx 10^{-4}$), high intensity and good collimation. Since the energy of the x-rays can be varied precisely and easily by a small rotation of a crystal monochromator, radiographs may be conveniently acquired at adjacent iodine-specific and iodine-nonspecific energies and then logarithmically subtracted. This subtraction substantially eliminates all the image contrast due to other body structures and thereby achieves maximum contrast in the visualization of the intra-arterial iodine. The method is referred to as dichromography.[1] Digital radiographs are acquired at x-ray energies slightly above and slightly below that of the K-edge in iodine at 33.16KeV. The iodine absorption cross section increases abruptly by a factor of six at this edge while the cross section of other elements remains relatively unchanged. The subsequent subtraction of these two images gives exquisite sensitivity to iodine alone. If this sensitivity is realized, it offers the prospect of arteriography and in particular, coronary arteriography by peripheral venous injection and therefore reduced concentration of iodinated compounds in the arteries. This would eliminate the need for catheterization and substantially reduce both the risk and the cost of the procedure.

Figure 1 shows how synchrotron radiation might be used with a multi-element Si(Li) detector to produce high resolution images of human arteries.

It is envisaged that in the final system the heart will be scanned using an x-ray beam which is 150mm wide and 0.6mm high. A line-scan procedure will be

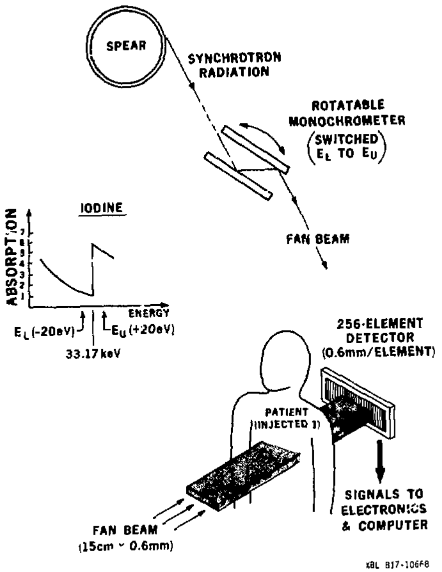

Fig. 1. Schematic layout of system for performing digital angiography using synchrotron radiation.

used in which the transmitted flux in each 0.6mm of the beam will be individually measured using a 256-element Si(Li) detector which will have an active volume 150mm long, 4mm high and 5mm thick and an element spacing of 0.6mm.

The imaging procedure will begin with the venous injection of about 40ml of Renografin 76*, an iodine containing compound commonly used in medical imaging. After about 12 seconds when the Renografin* has reached the coronary arteries the scan will start. The measurement interval will be 5ms following which the monochromator will be rotated 0.001° to the other energy for the second measurement. After each pair of measurements, the patient will be translated vertically by 0.6mm and the next pair of measurements performed. Since the iodine is rapidly dispersed throughout the body, the complete measurement must be accomplished within several seconds. We estimate that a complete scan of a 150mm x 150mm area can be completed within four seconds. The patient will receive a radiation dose of around 300mrad during the scan.

A silicon detector was chosen as the flux detector for this experiment for several reasons. A 5mm thick silicon detector has an efficiency of 70% at 33KeV and can be fabricated as a position-sensitive detector by replacing one of the contacts with a linear array of contacts connected to separate electronic readout channels. In a silicon solid-state detector, a photoelectric event produces one electron-hole pair for each 3.6eV of incident energy. A 33KeV photon therefore creates approximately 9,000 electron-hole pairs or a free charge 1.5×10^{-15}C. At this photon energy a flux of $10^7 sec^{-1}$ produces a current of 15nA. By measuring the current in each element the flux profile can be conveniently determined. Also inherent in the operation of such semiconductor detectors is the excellent linearity, stability and large dynamic range of the x-ray response. These properties are very useful in angiography where the flux per pixel can vary greatly because of tissue and bone variations but the flux difference must be measured to better than 0.5% if arteries are to be seen clearly. The charge transit time for a 5mm device is less than 1μsec so the response time is fast compared to the measurement time of several msec. This is important in angiography where the image must be acquired rapidly to reduce the effects of organ movement. An additional benefit of having all elements fabricated from the same silicon wafer is that the device gain of all elements is constant so the system can be calibrated accurately and conveniently.

Before embarking on the development of this large detector, we have been using the prototype detector in one of the wiggler x-ray beam lines at the Stanford Synchrotron Radiation Laboratory (SSRL) to study the feasibility of this concept.

Detector Characteristics and Fabrication

The prototype detector consists of a linear array of 30-elements which are 0.9mm x 7mm and have a center-to-center spacing of 1 mm. The geometry of the device and the electronics associated with each channel is shown in Fig. 2.

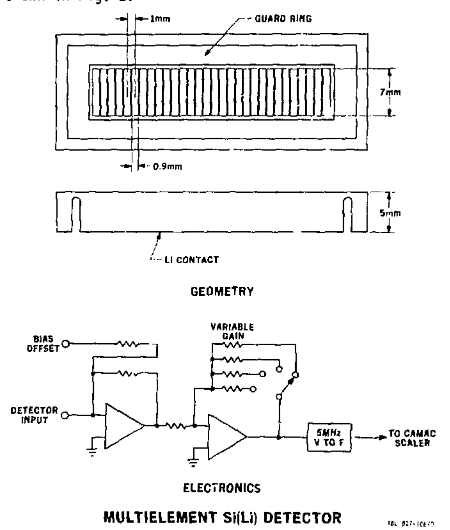

Fig. 2. Geometry of 30-element Si(Li) detector and the elecronics associated with each element.

The device was first fabricated as a 15mm x 39mm, 5mm thick Si(Li) grooved detector using standard procedures.[2] The rear contact of the detector is a common Li-diffused contact with a 4mm deep groove around the active area. Once the device had been tested as a single detector, the continuous gold contact was removed and the strip geometry of 0.9mm x 7mm pads with 1mm center-to-center spacing surrounded by a guard ring[3] was made by evaporating gold through a metal mask. The thickness of the gold contact is about 0.03μm and that of the Li contact is about 125μm. The purpose of the guard ring is to absorb the surface leakage current and to provide a good electric field profile in the pad region. With this geometry, the total device leakage current at room temperature is 6.2μA at an operating voltage of 600 volts. The expected leakage current, I_D, of each element at room temperature due to thermal generation is given approximately by:[4]

$$I_D = q\, n_i\, V / 2\tau_p$$

where q is 1.6×10^{-19} coulombs; n_i, the intrinsic carrier concentration, is 1.6×10^{10} carriers/cm^3; V, the depletion volume, is 0.035cm^{-3}; and τ_p, the minority carrier lifetime, is 2×10^{-3}sec. Since the calculated element leakage current is 22nA and the average leakage current is 10nA, the leakage current is primarily thermally generated and not surface leakage. The thermally generated current can be expected to decrease by about a factor of two for every 10°C decrease of the device temperature. We are currently fabricating a thermoelectrically cooled cryostat for the device which will cool the detector to -20°C and should reduce the leakage current to less than 1nA/element.

The resistance in parallel from each element to its neighboring elements and the guard ring was found to be 70K ± 15K ohms. This resistance was determined by measuring the leakage current of one element at the operating voltage as a function of a small differential voltage applied to the surrounding elements and the guard ring.

Electrical contact is made to each element using an array of Be-Cu spring contacts mounted on a circuit card. Figure 3 shows the detector mounted on the circuit card. The signals are carried from the detector to the electronic readout modules outside the x-ray beam enclosure using two 20-conductor ribbon coaxial cables with mass-terminated connectors.

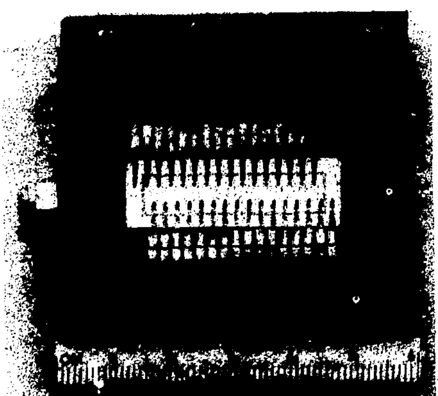

Fig. 3. Photograph of the 30-element Si(Li) detector with spring contacts in place.

The fabrication of the larger detector will be a challenging project. We currently plan to fabricate it from a single piece of silicon 170mm long, 12mm wide and 5mm thick. The device will have a similar groove and guard-ring structure as the initial detector. The element width, however, will be 0.5mm with a center-to-center spacing of 0.6mm. A photographic mask will be used to reduce the pad area and therefore minimize the leakage current and increase the inter-element resistance. Wire bonding to the detector elements may be used instead of spring contacts to improve the reliability of the 256 contacts.

Detector Electronics

The readout for this system requires the accurate measurement of the current flowing in each element. At SSRL a flux of 10^9 per second per mm^2 is available on the wiggler beam lines when the storage ring is operating in a dedicated mode at 3GeV and 60mA. Assuming that only 0.5% of the incident flux is transmitted through the patient, one estimates a photon flux of approximately 5×10^6sec^{-1} in each mm^2 and therefore a generated current of 7.5nA in each element of the prototype detector.

Since the detector inter-element resistance is relatively low (~70K ohms), it is important to use a high quality operational amplifier (OP-07) in the first stage of the electronics to keep the electronic noise and drift small. Provision is made in the electronics to change the gain and to bias off the detector leakage current from the data acquisition computer. A 5MHz voltage-to-frequency (V/f) converter is used to produce a digital output which is proportional to the input current. This digital output has excellent linearity, high resolution and a large dynamic range which complement the properties of the Si(Li) detector very well. Gating the V/f convertors with a common timing pulse of the desired period provides a measurement of the total flux in each element during the gating interval. The accuracy which can be achieved with this system with different fluxes has been presented elsewhere.[5] A typical clinical iodine concentration obtained with venous injection produces a flux difference of more than 2% and we estimate that an accuracy of better than 0.5% can be achieved in 5msec with the available flux from the the Stanford Synchrotron Radiation Laboratory (SSRL) wiggler beam line when SSRL is operating at 3GeV and 100mA.

The digital outputs from the individual elements are connected to separate CAMAC scalers which are gated with a common programmable CAMAC timer. The control of the sample position is done with stepping motors through the same CAMAC crate. The overall data acquisition and analysis system has been described elsewhere.[5]

Results

Initial studies with the prototype detector of the feasibility of synchrotron radiation for digital subtraction angiography were performed at SSRL in the spring of 1981. Detailed reports of the samples studied and the of images obtained in this experiment have been given elsewhere.[5,6,7] Very good initial images were obtained which indicate that this is a very promising technique. Only a summary of these results as they relate to the characteristics of the multi-element detector will be given here.

Cross-talk between detector elements can degrade the spatial resolution of the image. This cross-talk was measured by using a 0.5mm slit which was moved across the device in 0.05mm steps and then measuring the current in pairs of adjacent pads. As can be seen in Fig. 4, it is very low between elements which indicates that both the charge collection in the detector is well defined to the nearest pad and that the electronic feedthrough has been successfully minimized.

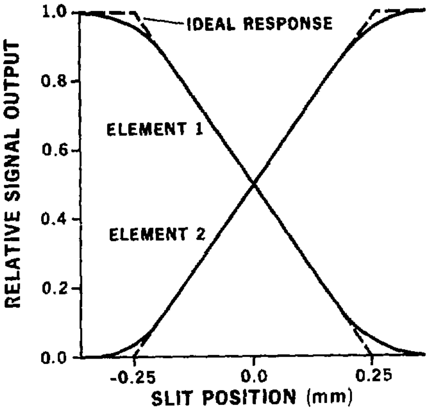

Fig. 4. Signal measured in two pads as a 0.5mm slit was scanned across the pads.

Images of both phantoms and excised animal hearts were obtained with this prototype detector. Due to limited availability at that stage of electronics, only eight adjacent channels of the detector were used to scan the heart in five adjacent vertical swaths each 48mm in height. A line-scan measurement (eight channels wide) was made, first 50eV above and then 50eV below the iodine K-edge before the object was moved vertically to the next raster position. An object area 48mm x 48mm was scanned, with 1mm x 1mm pixel size. The pedestals and gains of each electronic channel were monitored respectively by including 3mm lead absorbing strips at the top and bottom of each image and an unobstructed horizontal strip immediately above the lower lead strip. The total incident x-ray flux was monitored throughout by an upstream ionization chamber whose output was digitized and recorded with the Si(Li) detector data. This data was used to normalize each pixel to constant incident flux.

One of the set of phantom images, taken to demonstrate the strong, simultaneous suppression of contrast due to soft tissue and bone made possible with monoenergetic x-rays, is displayed in Fig. 5.

The figure shows that the energy resolution and dynamic range of the system are sufficient to eliminate the effects of bone and a variable amount of tissue in the difference image.

Figure 6 shows the arterial system of an excised calf heart obtained with two pixel sizes. Iodine was retained in the heart arteries by injecting into them a dental impression paste (CutterSil*) in which an appropriate concentration of NaI powder had been mixed. The heart was imaged for detector pixel sizes of both $1mm^2$ and $0.5mm^2$ to help select the width of the detector elements in the final detector. A pixel size of 0.5cm was achieved by placing a mask, with 0.5mm apertures spaced by 1mm and centered on the 0.9mm wide contacts, in front of the present detector and by acquiring two interlaced component images for each scan. The beam was also collimated to a matching 0.5mm height and the vertical motion after each line-scan was 0.5mm. Figure 6(a) and 6(b) show the difference images for $1mm^2$ and $0.5mm^2$ pixel size respectively. It is apparent that 1mm is marginally acceptable for clinical evaluation and that a resolution close to $0.5mm^2$ is preferable for clinical work. An element center-to-center spacing of 0.6mm has therefore been selected for the larger detector. With this spacing, a 15cm wide image can be obtained with 256 elements. This detector will give excellent image quality with a reasonable number of electronic readout channels.

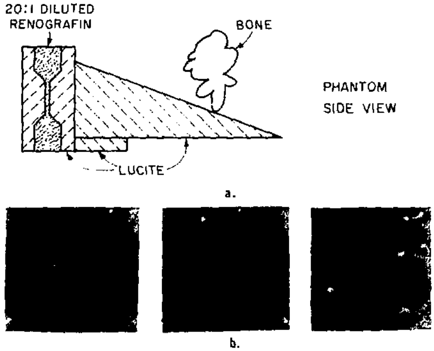

Fig. 5. Phantom image obtained with prototype system. 5(a) indicates the structure of the phantom. The row of images in 5(b) are, respectively, those obtained for the phantom above the K-edge, below the K-edge, and from their logarithmic subtraction. The iodine concentration was $2.5mg/cm^2$ in the 1.5mm diameter constricted lucite channel. The lucite wedge simulates a large tissue thickness variation and the bone is a human vertebral bone.

The images in Figure 6 inadvertently reveal arterial structure that is not really present. Towards the end of the circumflex artery (the major horizontal vessel), there appears a region about 1cm in length with lower than expected iodine concentration. Dissection of this region revealed an air bubble in the CutterSil*-NaI paste. Other smaller bubbles are evident in the left main descending artery. These air bubbles are easily preventable, but their occurrence accidentally in these images was useful from the standpoint of confirming that disease-like arterial structure could be seen with this system.

Summary

Multi-element Si(Li) detectors provide a method of making high precision x-ray flux measurements with good spatial resolution. These devices have excellent stability, efficiency, linearity and dynamic range. Techniques are available to fabricate them with large areas and complex contact geometries. They are therefore very promising detectors for many experiments using the intense x-ray fluxes which are available at synchrotron radiation facilities.

The electronic readout of the current in each element using a high quality preamplifier and voltage-to-frequency converter can provide an accurate flux measurement for many elements at reasonable cost.

Development of a 15cm wide wiggler x-ray beam line at a synchrotron radiation facility in combination with a 256-element Si(Li) detector may make possible the high-speed imaging of human arteries (especially the coronary arteries) non-invasively. With such a powerful instrument it may be possible to reduce the mortality of the present invasive procedure and to provide a diagnostic tool for the study and prevention of arterial disease.

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*Reference to a company or product name does not imply approval or recommendation of the product by the University of California or D.O.E. to the exclusion of others that may be suitable.

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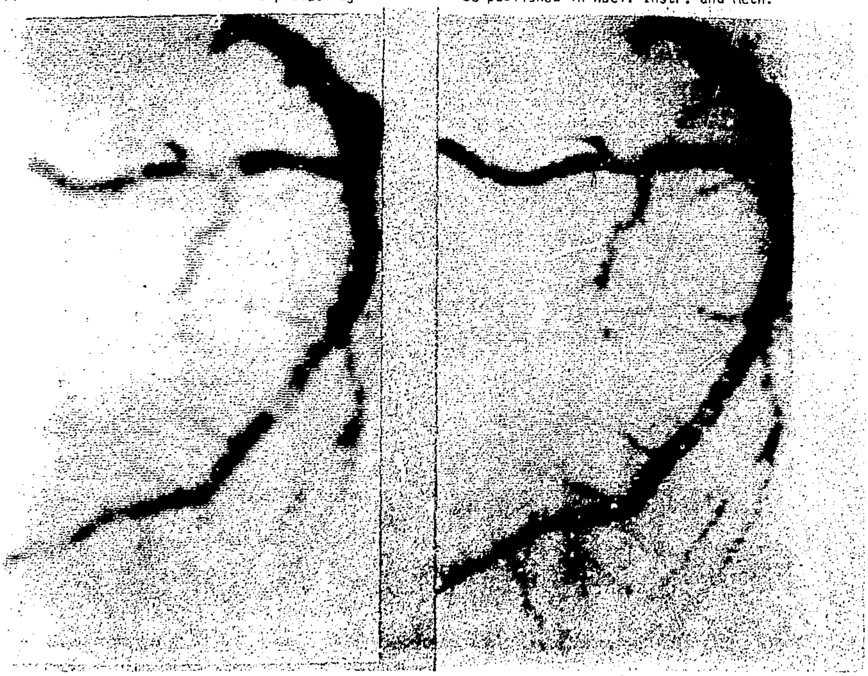

a. b.

Fig. 6. Difference calf heart images taken with (a) $1mm^2$ pixel size, and (b) $0.5mm^2$ pixel size.