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THE USE OF MULTIELEMENT DETECTOR SYSTEMS WITH
SYNCHROTRON X-RAY SOURCES

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ABSTRACT

The extremely high intensity and pulsed structure of synchrotron radiation x-ray sources put very demanding requirements on associated x-ray detectors. In current detector systems, trade-offs must be made between the efficiency, energy resolution, counting rate capability and the spatial resolution. Two detector systems are described which illustrate the optimization of these parameters for different applications of synchrotron radiation. One system is a segmented 16 channel multiwire proportional chamber which is used for fluorescent EXAFS measurements. The other is a 30 element Si(Li) linear detector array which is used for digital angiography experiments. The characteristics of these systems are discussed and recent results obtained with them are presented.

INTRODUCTION

The unique characteristics of synchrotron radiation x-ray sources have made possible many interesting and fruitful areas of research in physics, chemistry and biology. Although the mechanics of the measurements vary, generally they involve the detection of transmitted, scattered or fluorescent x-rays. These x-rays are often present in very high fluxes, highly collimated and monochromatized to a high energy resolution. Furthermore, they can reflect the complex time structure of the stored electron beam from which they are derived.

All of these properties present demanding requirements on x-ray detection systems. The ideal detector and its associated electronics would have an energy resolution comparable to that of the beam, would exhibit a fast response time and corresponding high-counting rate capability and would be capable of encoding the location at which the event is detected. However, current detectors do not possess all of these desirable characteristics simultaneously but emphasize some of them at the expense of others. Gas ionization chambers measure high fluxes with little energy, position or time information. Semiconductor detector spectrometers have moderately good energy resolution but limited rate handling capability. Scintillation detectors have good counting rate capability but limited energy and position discrimination. Multiwire proportional chambers can be designed with excellent spatial resolution and rate capabilities but only limited energy resolution. Other systems have similar limitations.

We have developed two types of multichannel detectors for use with synchrotron x-ray beams. Each system exhibits a specific optimization in characteristics which is tailored to the type of experiment being performed. Both detectors are segmented into elements which are coupled to separate readout channels to help distribute the counting rate and provide some spatial information. One detector is a segmented, multiwire proportional chamber which is used to obtain high-counting rate data from fluorescent extended x-ray absorption fine structure (EXAFS) experiments on dilute samples. Its major advantages are excellent counting rate capability, a large solid angle and moderate position and energy resolution. The second detector is a small 30 element Si(Li) detector which is used to achieve high speed flux measurements with good spatial resolution and high

efficiency in an experiment using synchrotron radiation for digital angiography.

SEGMENTED MULTIWIRE PROPORTIONAL CHAMBER

Fluorescent EXAFS is a sensitive method of determining the radial distance parameters of the atomic constituents of many substances. In many biological systems it is often interesting to study elements which are present in only parts per million concentrations in samples of only a few grams. For these studies and other dilute systems, the detection of the fluorescence x-rays from the sample is the most sensitive method of EXAFS measurement. We have developed a segmented, multiwire gas proportional chamber as a general purpose detector for fluorescent EXAFS experiments. The detector has a large solid-angle, good detection efficiency and excellent counting rate performance. Additional advantages of this detector include relatively simple construction, inexpensive, commercially available electronic readout and the ability to improve the signal-to-noise ratio by separating events emitted near 90° to the sample from those at other angles.

The detector geometry is shown in Figure 1. The active detector area is 190 cm x 190 cm and the thickness is 7 mm. It is filled with an Ar-CH₄ (93% - 7%) gas mixture. With this gas the detection efficiency at 6 keV is 80% while at 15 keV it drops to 11%. Other gases could be used to improve the efficiency at higher energies. The first 21 mm of the chamber is used as a drift region in which the charge produced by the photoelectrons is drifted to the multiplication region at the rear by a field of 40 V/mm. In addition to improving the efficiency of the detector, this region also serves to spread out the time distribution of the incident x-rays so that more than one event can be counted by each segment for each beam pulse giving excellent count rate performance. The multiplication region of the chamber is separated from the drift region by a grid of copper-beryllium wires. Proportional charge multiplication takes place near one of the anode wires in the anode plane. This plane consists of 20 μ m diameter gold-covered tantalum wire which are spaced 2 mm apart and are biased at +4200 V.

The information in the chamber is read out using the rear cathode plane. This electrode is segmented into a 4 x 4 array of 46 mm square pads which are connected to individual pulse discriminators. The electronic readout uses a Lecroy 7790* 16 channel discriminator board to produce a separate output rate from each pad. This board provides a single threshold to set the trigger level of all 16 discriminators. The output pulse width of all channels is also set with one control. If the input impedance of all channels is carefully adjusted to provide equal gain in all channels, the set-up of the detector is the simple adjustment of one threshold level. Some energy resolution can be achieved with this detector if the threshold is set to trigger just below the pulse height of the fluorescent x-ray energy. Each channel has a counting rate capability of about 1 MHz so that the total rate capability of the detector is 16 MHz. The output of the discriminator board is interfaced to the data acquisition computer with CAMAC scalars. The data acquisition program is a modified version of an earlier LBL program⁽¹⁾ which separately stores the 16 channels of fluorescent information and the transmission data from the ionization detectors.

The detector is mounted about 15 cm from the sample with a scattering shield in front to limit the field of view of the detector to sample area. A Z-1 x-ray filter and collimator assembly as developed by Stern et al⁽²⁾ is usually mounted between the sample and the detector to reduce the scattered x-ray intensity relative to the fluorescent x-ray signal.

An additional advantage of this detector arises from the spatial separation of the inner pads from the outer ones. This feature is useful since the polarization of the incident beam reduces the scattering at 90° resulting in a lower background in the central pads compared to the outer ones. A hexagonal pad structure could be used to further improve the signal in the central elements.

Examples of the spectra which have been obtained with this detector are shown in Fig. 2. These spectra are from a 1.6 $\mu\text{g}/\text{cc}$ concentration of Fe in a biological sample and the acquisition time per point in the edge region is one second. The incident x-ray flux on the sample was

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

approximately 5×10^{12} photons per second from an x-ray beam line at the Stanford Synchrotron Radiation Laboratory (SSRL). As can be seen from the figure, the data from a central pad are of considerably better quality than that from the outer pads although there is useful information in each spectrum. Data from all pads is usually combined using appropriate weighting factors before detailed EXAFS analysis.

MULTIELEMENT SILICON DETECTOR

The other detector is a segmented Si(Li) detector which is used in an experiment to study the application of synchrotron radiation to non-invasive x-ray imaging of coronary arteries. In this application the main detector requirement is good spatial resolution combined with high detection efficiency for 33 keV x-rays. The principle of the measurement, which is outlined in Fig. 3, consists of the measurement of the transmitted x-ray flux for energies immediately above and below the iodine K absorption edge at 33.169 keV. Since the iodine absorption cross section increases abruptly by a factor of six at this edge while the cross section of other elements is relatively unchanged, the difference of the two measurements will reflect only the iodine concentration and not the differences due to bone and tissue. For the imaging of coronary arteries, the patient would be injected in a vein with an iodine solution which would then flow through the coronary arteries. Since the iodine is rapidly dispersed throughout the body the complete measurement must be accomplished in less than 10 seconds. It is envisaged that in the final system, the heart will be scanned using an x-ray beam which is 150 mm wide and 0.6 mm high. The transmitted flux in each 0.6 mm of the beam will be individually measured using a 250 element Si(Li) detector which has an active volume 150 mm long, 4 mm high and 5 mm thick.

A silicon detector was chosen for this experiment for several reasons. A 5 mm thick silicon detector has an efficiency of 70% at 33 keV and is relatively easy to fabricate as a position-sensitive detector. This is done by replacing one of the contacts with a linear array of contacts which can be connected to separate electronic readout channels.

Inherent in the operation of such detectors is the excellent linearity and stability of its response to x-rays which results in a convenient and reliable calibration of the imaging system. The measurement interval will be 15 ms following which the monochromator will be rapidly rotated to the other energy for the second measurement. The blurring of the image by motion of the heart will be minimized using this procedure. After each pair of measurements, the patient will be translated vertically by 0.6 mm and the next pair of measurements performed. We estimate that a complete scan of a 150 mm x 150 mm area can be completed in less than eight seconds.

Before embarking on the development of this large detector and the beam line necessary for clinical tests, we have been working with the Hughes-Hofstadter group at Stanford University to study the feasibility of this concept with a smaller Si(Li) detector using one of the wiggler beam lines at SSRL. This detector, which is illustrated in Fig. 4, consists of a linear array of 30 elements which are 0.9 mm x 7 mm and have a center-to-center spacing of 1 mm. It is used with a beam which is collimated to have a vertical height of 1 mm to give a spatial resolution of 1 mm^2 in the final image. The rear contact of the detector has a common Li diffused contact with a 3 mm deep groove around the active area to reduce the surface leakage current. The pattern of the front contact was made by a gold evaporation through a metal mask. The 30 elements are surrounded by a guard ring which is grounded to minimize the leakage current in each element. The detector is currently operated at room temperature at a bias of 600 V and has an average leakage current in each element of 10 nA. This leakage current can easily be reduced by cooling the detector since it is primarily from thermally generated charge in the active volume.

The readout for this system requires the accurate measurement of the current flowing in each element. In a silicon solid-state detector, a photoelectric event produces one electron-hole pair for each 2.98 eV of incident energy. A 33 keV photon, therefore, creates approximately 11,000 electron-hole pairs or a free charge of $1.8 \times 10^{-15} \text{ C}$. A photon flux of 10^7 per second therefore produces a current of 18 nA. 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 3 GeV and 60 mA. Assuming that only 1% of the incident flux is transmitted through the patient,

one estimates a flux of approximately 10^7 in each pixel.

The readout electronics, shown in Fig. 4, employs a high quality amplifier followed by a second amplifier to bias off the leakage current and provide variable gain. A 5 MHz voltage-to-frequency converter is then used to produce a digital output which is proportional to the input current. These outputs are then connected to CAMAC scalars which are gated with a programable CAMAC timer.

This detector has been used in several preliminary measurements at SSRL using one of the wiggler beam lines. Using only eight contiguous channels of the detector due to electronic limitations, we have studied a variety of samples by taking the data in swaths and gradually accumulating the complete image. A plastic phantom with a 1 mm channel of iodinated water with a vertebrae bone in front was scanned using a measurement time at each energy of 16 msec per line segment (10^8 photons/pixel). An excellent image of only the iodine channel was obtained when the image above the edge was divided by the image below the edge (ratio image). A second measurement was made using a calf's heart in which the arteries were filled with a Cuttersil (a dental paste) which had been loaded with a realistic clinical concentration of iodine (18 mg/cm^3) which gives an 8% difference signal for a 1.5 mm artery. Two pairs of images were taken to evaluate the resolution necessary for diagnosis. In the first pair, the heart was scanned with a pixel size of 1 mm^2 and in the second a 0.5 mm set of lead slits was placed in front of the detector and the beam height was reduced to 0.5 mm so that the pixel size was reduced to 0.5 mm^2 . Figure 5(a) shows the 0.5 mm^2 image taken below the iodine edge and Fig. 5(b) is the same image taken just above the edge. Figures 5(c) and 5(d) show the final ratio images that were achieved with 1 mm^2 (5c) and 0.5 mm^2 (5d) pixel size. In both ratio images the arteries are clearly seen but there is significantly better resolution with the 0.5 mm^2 pixel size. Examination of the calf's heart following the experiment verified that the two voids visible in the image were due to air bubbles and that some of the graininess of the image was due to incomplete mixing of the iodine in the Cuttersil. We are currently developing mechanical and electronic instrumentation to enable us to take images within several seconds with this prototype detector so that we can study the effect of sample movement using animate samples.

SUMMARY

Optimization of device parameters for different applications can produce x-ray detectors which operate well with synchrotron x-ray sources. A segmented, multiwire proportional chamber has been built which provides excellent count rate performance, some position information and good solid angle for fluorescent EXAFS experiments. For digital angiography experiments the detection requirements are quite different. A multielement Si(Li) detector has been developed for this experiment which gives good spatial resolution, stable operation and excellent linearity. Development of a larger detector system with an appropriate x-ray beam line may provide high resolution images of coronary arteries non-invasively. These detector systems illustrate that careful design can result in very effective utilization of synchrotron x-ray sources.

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FIGURES

Fig. 1. Geometry of segmented multiwire proportional chamber.

Fig. 2. EXAFS spectra from 1.6 $\mu\text{g}/\text{cc}$ Fe biological sample. 2(a) is from a corner detector element, (b) is from a central element and (c) is from an element on the lower edge.

Fig. 3. Outline of method for digital imaging of coronary arteries.

Fig. 4. Geometry of 30 element Si(Li) detector and the electronics associated with each element.

Fig. 5. Images of calf's heart taken with arteries filled with iodine-loaded Cuttersil. Figure 5(a) was taken below the iodine edge with 0.5 mm^2 pixel size and Fig. 5(b) is the same image taken above the iodine edge. Figure 5(c) is the ratio image (above edge image divided by below edge image) obtained with 1 mm^2 pixel size and Fig. 5(d) is the same image obtained with 0.5 mm^2 pixel size.

SEGMENTED MULTIWIRE PROPORTIONAL CHAMBER

XBL 817-10671

Fig. 1. Geometry of segmented multi-wire proportional chamber.

XBL 817-10672

Figure 2. EXAFS spectra from 1.6 $\mu\text{g}/\text{cc}$ Fe biological sample. 2(a) is from a corner detector element, (b) is from a central element and (c) is from an element on the lower edge.

XBL 817-1066a

Figure 3. Outline of method for digital imaging of coronary arteries.

GEOMETRY

ELECTRONICS

MULTIELEMENT Si(Li) DETECTOR

LBL 817-10670

Figure 4. Geometry of 30 element Si(Li) detector and the electronics associated with each element.

XBB 817-6988

Figure 5.