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X-ray imaging with a silicon microstrip detector coupled to the RX64 ASIC

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Abstract

A single photon counting X-ray imaging system, with possible applications to dual energy mammography and angiography, is presented. A silicon microstrip detector with 100 μm pitch strips is coupled to RX64 ASICs, each of them including 64 channels of preamplifier, shaper, discriminator and scaler. The system has low noise, good spatial resolution and high counting rate capability. Results on energy resolution have been obtained with a fluorescence source and quasi-monochromatic X-rays beams. Preliminary images obtained with an angiographic phantom are presented.

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Keywords: X-rays; Imaging; Angiography; Silicon detector; ASIC

1. Introduction and experimental setup

In this paper we report on the development of an X-ray imaging system with single photon counting capability based on silicon strip detectors and on the RX64 readout ASIC [1], which is an evolution of the previous RX32 and COUNT32 ASICs [2]. Silicon microstrip detectors with adequate spatial resolution are well suited for digital X-ray imaging, where a two-dimensional image is obtained from the one-dimensional array of strips by scanning the object along the

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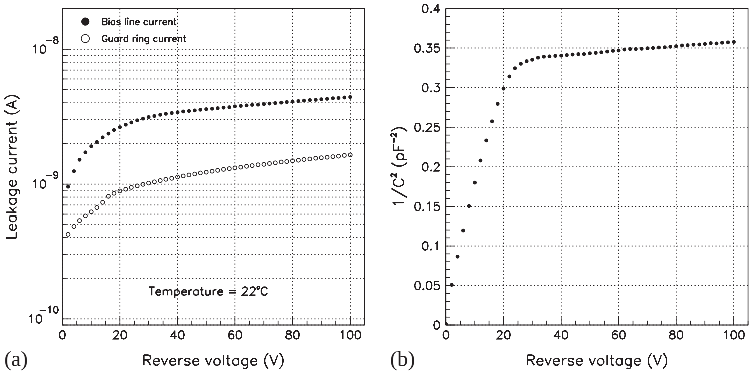

Fig. 1. Results of $I-V$ and $C-V$ measurements: (a) bias line and guard ring current vs. reverse voltage; (b) $C-V$ measurement for a group of four contiguous strips.

orthogonal direction. For specific diagnostic applications, such as mammography and angiography at the iodine K-edge, the dual energy technique makes it possible to isolate materials of specific interest via the energy subtraction method, providing an enhancement of the image contrast. Clinical application has so far been limited by the broad energy spectrum of conventional X-ray sources, with the exception of synchrotron radiation [3,4], which however, is available only in a few locations. Using a quasi-monochromatic X-ray beam [5] we have characterized two prototype silicon detectors with 128 and 384, respectively, equipped channels.

We have used 132 and 400 strip silicon detectors[1] with strip length of 1 cm and pitch of 100 μm. Detectors have been characterized with $I-V$ and $C-V$ measurements under the Alessi probe station REL 4500 using the HP4145B semiconductor parameter analyzer and the HP4284A precision LCR meter.

In Fig. 1a the bias line current and guard ring current as a function of the reverse voltage for a 132-strip detector are shown. A typical measurement of capacitance vs. reverse voltage for a group of four contiguous strips of a 400-strip detector is displayed in Fig. 1b. In order to maintain a good noise performance, both the capacitance and the leakage current per strip must be kept as low as possible. A practical limit for our electronics (see below) is 5 pF and 100 pA, respectively. With a typical full depletion voltage of 25 V, at the operating voltage of 60 V the leakage current is between 28 and 44 pA/strip.

The efficiency for converting X-rays via the photoelectric effect must be high enough. With microstrips orthogonal to the incoming beam (front configuration), the quantum efficiency of a 300 μm Si-wafer for X-rays of medical energies is at best a few percent; a good detecting efficiency can be achieved by orienting the microstrips parallel to the incoming beam (edge-on configuration), as shown in Fig. 2 (left). Simple calculations of efficiency in the two configurations, which take into account all absorption processes in the layers located upstream from the detector (70 μm of Al in front configuration, 765 μm of insensitive Si in edge-on configuration) and only photoelectric conversion in the active Si volume (300 μm and 10 mm in front and edge-on configurations, respectively), lead to the results shown in Fig. 2 (right). In the whole interesting energy range for angiography and mammography (18−36 keV) the silicon detector is much more efficient in the edge-on configuration.

Our readout electronics has a binary architecture, i.e. 1-bit (yes/no) information per strip, suitable for high counting rate applications. The readout chain is based on the RX64 ASIC, designed and manufactured in the AMS 0.8 μm technology: it consists of 64 channels, which for the analog part contain three basic blocks: charge preamplifier, shaper and discriminator. A set of 64

[1] Detectors were designed and built by ITC-IRST, Trento, Italy.

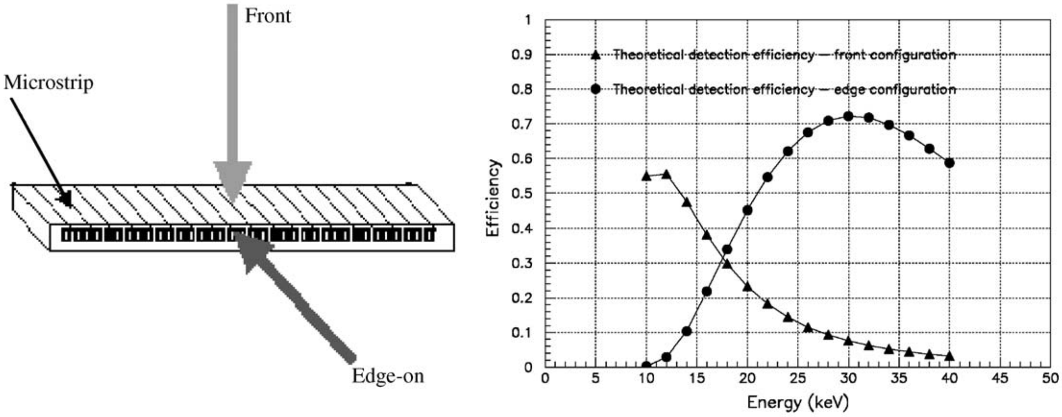

Fig. 2. Front and edge-on configurations (left panel); quantum efficiency in front and edge-on configurations vs. photon energy (right panel).

20-bit asynchronous counters is integrated in the chip. The main characteristics of the RX64 chip are: tunable gain (60–105 μV/electron) and peaking time (500–1000 ns), a noise (ENC) of 167e RMS ($C_{det} = 2.5$ pF, $I_{det} = 100$ pA), and a counting rate of 200 kHz/channel. An internal calibration circuit, also integrated in the RX64, is used to apply to the inputs of the chip test pulses of known charge by means of capacitors of nominal value 75 fF. The microstrip detector and the RX64 chips are glued on a multilayer PCB which contains the signal and power lines and serves also as a mechanical support. Data acquisition is done with PCI-DIO-96 or DAQCard-DIO-24 digital I/O cards (National Instruments) controlled by a program written under LabVIEW 6i.

2. Energy calibration

The analog parameters of the RX64 ASIC can be obtained with purely digital control by scanning the discriminator threshold (set via a DAC built inside the RX64) while applying a given number of test pulses to the input. After amplification and shaping we measure the fraction of pulses which exceeds the discriminator threshold; by differentiating the measured counts we obtain a Gaussian distribution whose σ and peak position are equal respectively to the RMS value of noise and to the signal amplitude at the discriminator input. The final result is shown in Fig. 3 (left) for a single channel: one can see that the circuit is linear up to $\approx$8000 electrons (X-ray energy up to 30 keV). In this region, a linear fit to the data points allows to extract the gain and the comparator offset for each channel. The gain distribution for the 128 channels has an average value of 61.6 μV/e and a width of 1.4 μV/e with a small ($\approx$3.5%) systematic difference between the two RX64 chips. The RMS value of noise obtained by fitting the Gaussian function to the differential distributions is equal to 8.1 mV. Thus, the equivalent noise charge is equal to 131 electrons RMS, as expected from simulations. The distribution of comparator offsets has an RMS value of 3.2 mV, a factor of two smaller than the noise level, which is crucial since all 64 channels are operated with a common threshold. With the 384-channel prototype, in similar conditions of low gain as above (63.7 μV/e), the measured ENC was of 176 e RMS with the six RX64 ASICs alone, and of 184 e RMS after connecting the pitch adapter to the ASICs.

In order to eliminate the uncertainty on the injected charge, we calibrated the system using an X-ray source which consists of a primary 10 mCi (370 MBq) ^{241}Am source which excites characteristic X-rays in six different targets (Cu, Rb, Mo, Ag, Ba, Tb) mounted on a rotary holder, with K_{α} emission lines ranging from 8.04 to

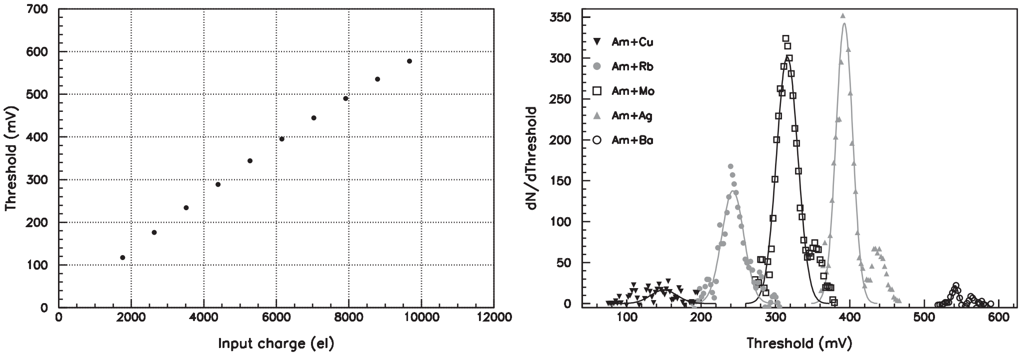

Fig. 3. Linearity of the amplifier and shaper part of the RX64 chip (left panel); Gaussian peaks obtained with the ^{241}Am source and five different targets. Gaussian fits to the K_α peak are superimposed (right panel).

44.23 keV. Measurements have been taken with the strips oriented orthogonally to the incoming photons and biased at 60 V. The distributions obtained by differentiating the measured counts are shown in Fig. 3 (right) for a typical strip together with the results of Gaussian fits to the K_α peak. The K_β peak for Ag, Mo and Ba is clearly visible.

The gain is evaluated by means of a linear fit to the Gaussian peak threshold values vs. energy (see Fig. 3 right) in the region where the response of the chip is linear (i.e. below 30 keV, as obtained above). The average gain extracted from the Am source data is equal to 17.0 mV/keV, which corresponds to 61.7 μV/e (3.63 eV per electron–hole pair in Si), quite compatible with the one extracted with internal calibration. The RMS of the gain distribution is 0.3 mV/keV with a small ($\approx$2.7%) systematic difference between the gain of the two RX64 chips. The RMS value of noise has been estimated from the Ag data for which the K_α and K_β peaks are well separated, obtaining 11.22 mV (ENC$\approx$180 e RMS, which includes fluctuations of the number of electron–hole pairs).

The experimental setup with quasi-monochromatic X-rays is essentially the same of Ref. [5,6] and is shown in Fig. 4. X-rays from an ordinary tube are selected in energy by means of a double collimator and a highly oriented pyrolithic graphite mosaic crystal; the average energy may be changed by two remotely controlled isocentric goniometers. The X-ray transmission through a test object is measured by the silicon microstrip detector, with a resolution of 100 μm along the vertical direction. Measurements have been taken using two different detector orientations: microstrips orthogonal to the beam (front configuration) or parallel to the beam (edge-on configuration): results are shown in Fig. 5 (left). The peak threshold values obtained at different beam energies are shown in Fig. 5 (right) for both detector orientations, together with the results obtained with the ^{241}Am source. The agreement between the different data sets is very good, demonstrating the quality of the tuning of the quasi-monochromatic X-ray beam. The Gaussian width does not depend on photon energy and has an average value of 13.4 mV; its increase with respect to the one obtained with the ^{241}Am source reflects the energy spread of the quasi-monochromatic beam [6].

3. Imaging of an angiographic test object

In the conventional hospital procedure, an iodate contrast medium (with a typical concentration of 370 mg/ml) is injected directly into the coronary arterial system via a long catheter fitted in the femoral artery while X-ray transmission images are taken. A less invasive procedure requires injection into a peripheral vein with a much more diluted contrast medium, hence a higher image contrast is needed. The K-edge

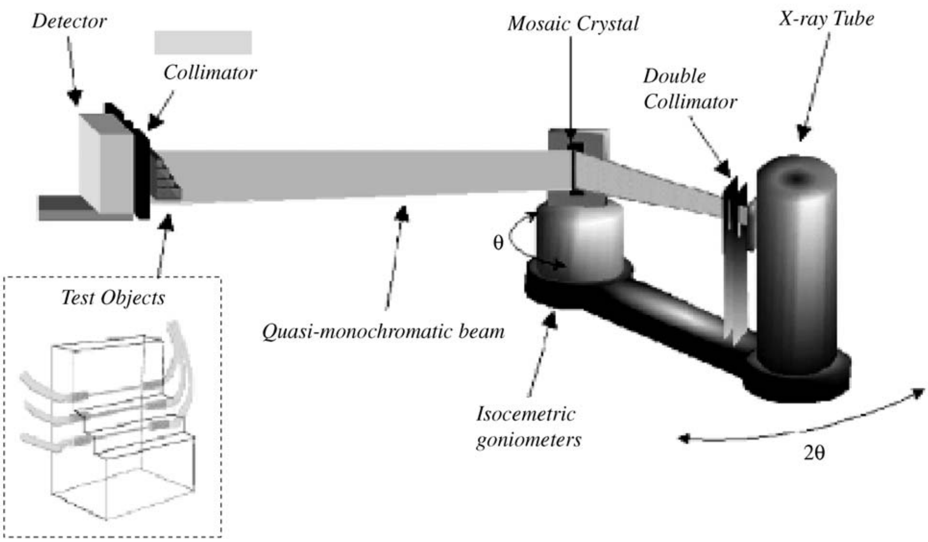

Fig. 4. The quasi-monochromatic X-ray beam setup.

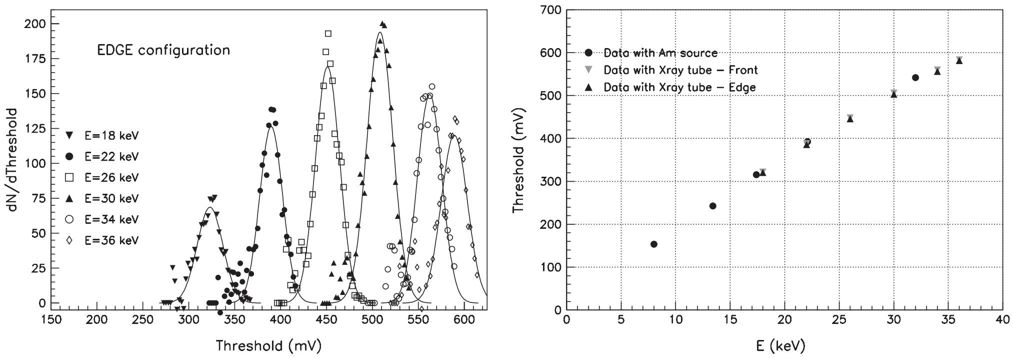

Fig. 5. Gaussian peaks obtained with X-ray beam at six energies (left panel); corresponding discriminator threshold values vs. energy for a typical strip, collected with Am source and X-ray beam (right panel).

subtraction angiography consists of two monochromatic photon beams, immediately below and above the iodine K-edge. Two images are taken simultaneously with digital detectors and then a digital (logarithmic) subtraction is performed pixel-by-pixel, thus extracting the iodine signal.

Our quasi-monochromatic X-ray beam with the 128-channel detector in the edge-on configuration has been used to acquire images of a test object at two different average energies (namely 31 and 35 keV) suitable for the K-edge subtraction. As a test object we used a plexiglass step wedge phantom, 40 mm high and 30 mm wide and with 10–45 mm thickness. Three cylindrical cavities of 1 mm diameter have been filled with iodate contrast medium of concentration 370 mg/ml. Images of the test object have been taken separately with the 31 and the 35 keV setting of the beam. The measured number of counts has been corrected for the X-ray beam intensity profile and for the different flux and detector efficiency at the two energies. The two-dimensional image is obtained by moving the object along the orthogonal direction. The image taken at 35 keV is shown

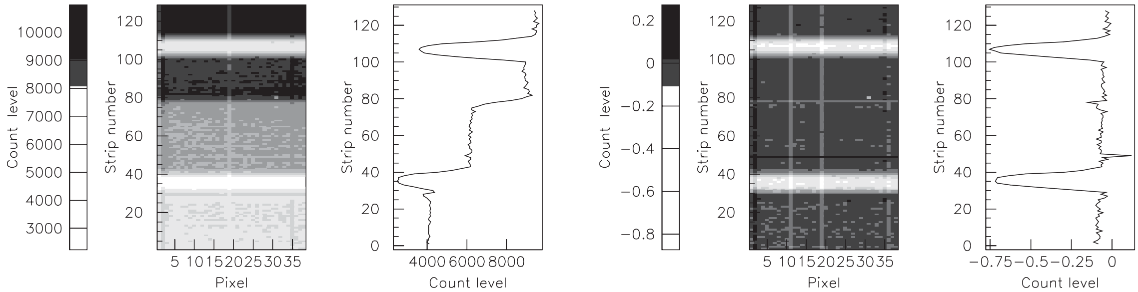

Fig. 6. Images and vertical profiles of the test object: count level at 35 keV (left panel), logarithm of the normalized ratio of counts at 31 and 35 keV (right panel).

in Fig. 6 (left), together with its mean vertical profile. We have observed that in the image at 31 keV (not shown) the measured detail contrast is about 2.5 times lower than the one measured at 35 keV; furthermore, in both images a vertical gradient due to the attenuation characteristics of the step-wedge phantom is present. The image obtained by performing a logarithmic subtraction of the images collected at 31 and 35 keV is shown in Fig. 6 (right) together with its mean vertical profile. It can be observed that no gradient due to the attenuation characteristics of the step wedge phantom is present and that the contrast generated by the iodate solution is very high. As a consequence of these two facts, the detail visibility is dramatically improved in the final image with respect to the primary images. This demonstrates the system's ability to image 1 mm vessels at standard iodate concentration and potentially at lower concentrations as well. Work is in progress towards exploiting the 384-channel detector and then implementing a larger field X-ray source and detector suitable for clinical angiography.

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