A silicon strip detector coupled to the RX64 ASIC for X-ray diagnostic imaging

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Abstract

First results from a silicon microstrip detector with 100 μm pitch coupled to the RX64 ASIC are presented. The system is capable of single photon counting in digital X-ray imaging, with possible applications to dual energy mammography and angiography. The main features of the detecting system are low noise, good spatial resolution and high counting rate capability. The energy resolution and the conversion efficiency of the system are discussed, based on results obtained with fluorescence X-ray sources and quasi-monochromatic X-ray beams in the 8–36 keV energy range, with strips being either orthogonal or parallel to the incoming X-rays. We present also preliminary imaging results obtained with a plexiglass phantom with tiny cylindrical cavities filled with iodate solution, simulating patient vessels; in this case the X-ray beam has two components, respectively below and above the iodine K-edge at 33.17 keV.

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1. Introduction

The present paper describes the development of a detecting system for position-sensitive X-ray measurement based on silicon strip detectors employed in the single photon counting mode and on the RX64 readout chip [1], which is
derived from the previous RX32 and COUNT32 ASICs [2].

Silicon microstrip detectors with good spatial resolution are well suited for digital X-ray imaging for diagnostic applications. A two-dimensional image is obtained from the one-dimensional array of strips by scanning the object along the orthogonal direction. The use of silicon detectors has the advantage of reducing the dose to the patient, with respect to traditional screen/film combination, without losing in image definition. Furthermore, all possibilities of processing and transferring digital data are guaranteed. In particular, for specific diagnostic applications, such as mammography and angiography at the iodine K-edge, the dual energy technique [3] makes it possible to isolate materials of specific interest via the energy subtraction method, providing an enhancement of the image contrast. The clinical application has been so far limited by the broad energy spectrum of conventional X-ray sources, with the exception of synchrotron radiation [4,5], which however is available only in a few locations.

A prototype with 128 equipped channels has been built and tested both with radioactive sources and with a quasi-monochromatic X-ray beam [6] based on an X-ray tube and an array of mosaic crystals. The apparatus consisting of quasi-monochromatic X-ray beam and silicon microstrip detectors is suitable for clinical application to dual energy diagnostic imaging.

2. Microstrip detector

For the first prototype we have used a 132-strip detector with FOXFET biasing. Each strip is 1 cm long and has a pitch of 100 μm. Detectors have been characterized with $I-V$ and $C-V$ measurements under a semiautomatic Alessi probe station REL 4500: $I-V$ measurements were obtained with the HP4145B semiconductor parameter analyzer and $C-V$ measurements with the HP4284A precision LCR meter. In Fig. 1a the bias line current and guard ring current as a function of the reverse voltage are shown. A typical measurement of capacitance vs. reverse voltage for a group of four contiguous strips is displayed in Fig. 1b.

The most relevant geometrical and electrical parameters of the microstrip detectors are summarized in Table 1. In order to maintain a good noise performance, both the capacitance and the leakage current per strip must be as low as possible.2

If the microstrips are oriented orthogonal to the incoming beam (front configuration), the efficiency of a 300 μm Si-wafer for converting X-rays of medical energies via the photoelectric effect is below 30%. A good detecting efficiency can be achieved by orienting the strips parallel to the

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1 Detectors were built by ITC-IRST, Trento, Italy.
2 Our electronics is optimized for detector capacitance of 1–3 pF per strip and leakage current below 100 pA per strip [7].
incoming beam (edge-on configuration) so as to have a long active length [8,9]. In this configuration, the insensitive region around the strips acts as an upstream absorbing layer and consequently its thickness must be kept as low as possible. Theoretical calculations of quantum efficiency in the two configurations have been carried on taking into account the absorption layers located upstream from the detector (70 µm Al foil used as light shield in front configuration, 765 µm of insensitive silicon in edge configuration) and the photoelectric probability in the active silicon (300 µm and 10 mm in front and edge configurations, respectively). Results are shown in Fig. 2 and indicate that in the whole interesting energy range for angiography and mammography (18–36 keV) the detector is much more efficient in the edge configuration.

Table 1
Summary of the main electrical parameters measured on the microstrip detectors

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Depth</td>
<td>300 µm</td>
</tr>
<tr>
<td>Strip length</td>
<td>10 mm</td>
</tr>
<tr>
<td>Channels</td>
<td>132</td>
</tr>
<tr>
<td>Strip pitch</td>
<td>100 µm</td>
</tr>
<tr>
<td>Insensitive region thickness</td>
<td>765 µm</td>
</tr>
<tr>
<td>Depletion voltage</td>
<td>≈ 25 V</td>
</tr>
<tr>
<td>Leakage current per strip at full depletion</td>
<td>≈ 35 pA</td>
</tr>
</tbody>
</table>

3. The front-end chip RX64

The readout electronics is based on a binary architecture, i.e. each channel of the front-end electronics provides 1-bit (yes/no) information in response to the signal generated in a given detector strip. Such an architecture is suitable for high counting rate applications, since the amount of data to be handled is minimized already in the front-end part [10,11].

The readout chain is based on the RX64 chip [1] which consists of 64 channels and is used to simultaneously process signals and store data from 64 strips of the silicon detector.

The analog part of each front-end channel contains three basic blocks: preamplifier, shaper, discriminator and is devoted to the treatment of the signal from one strip. A set of 64 counters is also integrated in the chip. The main characteristics of the RX64 chip [7,12] are summarized in Table 2.

The pre-amplifier stage is a charge sensitive amplifier which integrates the input signal from the strip. The preamplifier is followed by a shaper circuit providing noise filtering and semi-Gaussian pulse shaping with a peaking time tunable in the range 500–1000 ns. The analog front-end channel is ended with a continuous time discriminator. A differential scheme for setting the discriminator threshold allows the use of the chip either for positive or for negative input signals. Since the only practical solution is a common threshold applied to all channels in the chip, special attention was paid to matching problems in the discriminator design, in order to minimize the channel-to-channel offset spread. The digital information at the output of each discriminator

Table 2
Summary of the main parameters of the RX64 ASIC

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Measured value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gain</td>
<td>60–90 µV/electron</td>
</tr>
<tr>
<td>ENC (C_{det} = 2.5 pF, I_{det} = 100 pA)</td>
<td>167 el. RMS</td>
</tr>
<tr>
<td>Input range</td>
<td>up to 10,000 el.</td>
</tr>
<tr>
<td>Peaking time</td>
<td>500–1000 ns</td>
</tr>
<tr>
<td>Max. counting rate</td>
<td>100–200 kHz/channel</td>
</tr>
<tr>
<td>Power consumption</td>
<td>2.5 mW/channel</td>
</tr>
</tbody>
</table>

Fig. 2. Theoretical calculations of quantum efficiency for the microstrip detector in front and edge configurations as a function of photon energy.
is stored in a 20-bit asynchronous counter. Taking into account a fully parallel architecture of the readout system, including the counters which serve as memory buffers, the counting rate of the system is determined by the pile-up effects in the analogue part of the front-end electronics, and is on the level of 200 kHz per channel [2]. The data from 20-bit counter can be read out from the chip via 8-bit data bus with 10 MHz clock, so the time for transmitting the data of the chip is usually negligible.

An internal calibration circuit is also integrated in the RX64 IC and it is used to apply to the input of the chip a given number of test pulses of given amplitude by means of calibration capacitors of nominal value of 75 fF.

The 132-strip detector and the two RX64 chips needed for the readout are glued on a multilayer circuit which contains the signal and power lines and works also as a mechanical support. Data acquisition is done with PCI-DIO-96 or DAQ-Card-DIO-24 digital I/O cards (National Instruments) controlled by a program written under LabView 6i.

4. Measurements with RX64 internal calibration

The binary architecture implemented in the RX64 chip has certain implications on the testing methods which we have used because, for routine measurements, only the output of the scalers is available. The basic analog parameters—i.e. gain and noise—can be obtained by scanning the threshold of the discriminator while for each value of the threshold a given number of test pulses of given amplitude is applied to the input [7,12]. At the output we measure the fraction of pulses which after amplification and shaping exceeds the discriminator threshold and is counted by the scaler. If the amplitude distribution of noise is Gaussian, the measured number of counts vs. threshold is described by the error function. By differentiating the measured counts we obtain a Gaussian distribution whose sigma is equal to the RMS value of noise at the discriminator input. The detailed description of this procedure can be found in Ref. [2].

An example of measurement performed according to the procedure described above is shown in Fig. 3 for five different values of input charge in the case of detector biased at the depletion voltage and the front-end electronics working with a peaking time of ~500 ns. For threshold values below 80 mV we observe a rapid increase of counts due to noise pulses. As one can expect, the widths of amplitude distributions in Fig. 3b are independent of the input signal since these distributions are uniquely determined by the noise at the input of the comparator.

The discriminator threshold values corresponding to the peaks of these distributions (50% efficiency) determine the signal amplitudes at the discriminator input. The Gaussian peak threshold value as a function of the input charge for a typical channel is shown in Fig. 4 where it can be seen that the circuit is linear up to approximately 6000 electrons of input signal, which correspond to an X-ray energy range up to 20 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 of the 128 channels extracted from small input signals (below 5000 el.) has an
average value of 61.6 μV/el and a width of 1.4 μV/el. The RMS value of noise obtained by fitting the Gaussian function to the distributions shown in Fig. 3b is equal to 8.1 mV.\(^3\) Thus, the equivalent noise charge is equal to 131 electrons RMS\(^4\) in agreement with what expected from simulations [7]. One of the most critical factors for the RX64 chip is the spread of comparator offsets, since for all the 64 channels a common threshold is applied. The design goal was to keep the offset spread negligible with respect to the noise level. The measured distribution of offsets has an RMS value of 3.2 mV, which is a factor of 2 smaller than the noise level.

5. Measurements with Americium source

The value of the calibration capacitor depends on the processing parameters and it is not known accurately enough (the nominal value of 75 fF is guaranteed only with about 10% of accuracy) to be used for absolute calibration. In order to eliminate the uncertainty on the actually 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. The main characteristics of each target are summarized in Table 3.

Measurements have been taken with the strips oriented orthogonal to the incoming photons and biased at 60 V. The distributions obtained by differentiating the measured counts are shown in Fig. 5 for a typical strip together with the results of Gaussian fits to the K\(_a\) peak. The gain is evaluated by means of a linear fit to the Gaussian peak threshold values as a function of energy (see Fig. 6) in the energy range where the response of the chip is linear (i.e. below 30 keV). The average gain extracted from the Am source data is equal to 17.0 mV/keV, which corresponds to 61.7 μV/el, in excellent agreement with the value extracted with the internal calibration system. The RMS value of noise has been estimated from the Ag data for which the K\(_a\) and K\(_\beta\) peaks are well separated, obtaining 11.22 mV (corresponding to \(\approx 180\) ENC). The RMS noise value is higher than the one obtained with the internal calibration both because of the larger leakage current due to the higher bias voltage applied to the strips and due to the effects of comparator non-linearity.

6. Measurements with quasi-monochromatic X-ray beam

The system [6,13] is schematically depicted in Fig. 7. X-rays are produced by a tungsten anode.
X-ray tube and monochromatized by Bragg diffraction in a mosaic crystal. The X-ray tube is mounted on a goniometer, so that the desired Bragg angle can be selected. A double lead collimator is placed downstream of the tube so as to match the incident X-ray beam to the crystal size. The monochromator, placed at 250 mm from the X-ray tube focus, is a highly oriented pyrolytic graphite crystal (HOPG) with a thickness of 0.1 cm and a surface area of $2.8 \times 6.0 \, \text{cm}^2$.

The detector was placed at $\approx 90$ cm from the crystal. A 300 μm wide collimator was located immediately upstream from the detector. Measurements have been taken with the microstrips oriented both orthogonal (front configuration) and parallel (edge-on configuration) to the beam axis.

The Gaussian peak threshold values obtained by means of threshold scans at different beam energies are shown in Fig. 6 for both the detector

Fig. 5. Gaussian peaks obtained with the americium source and five different targets. The $K_b$ peak for Ag, Mo and Ba is clearly visible. Gaussian fits to the $K_a$ peak are superimposed.

Fig. 6. Discriminator threshold values corresponding to Gaussian peaks as a function of energy for a typical strip. Data collected with Am source and quasi-monochromatic X-ray beam are superimposed.

Fig. 7. Schematical representation of the experimental apparatus and of the step wedge plexiglass phantom used as a test object.
orientations together with the results obtained with the americium source. It can be seen that 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 extracted from the fits to the differentiated measured counts does not depend on photon energy and has an average value of 13.4 mV. The enlargement of the Gaussian width with respect to the one obtained from data collected with the americium source reflects the energy spread of the quasi-monochromatic beam [13].

7. Imaging test

The quasi-monochromatic X-ray beam with the detector in the edge-on configuration has been used to acquire images of a test object at two different energies (namely 31 and 35 keV) suitable for the K-edge subtraction angiography application.

Coronary angiography is a medical examination aimed at the visualization of coronary arteries. In the conventional hospital procedure, an iodate contrast medium (with a typical concentration of 370 mg/ml of iodine) is injected directly into the coronary arterial system via a long catheter fitted in the femoral artery while an X-ray transmission image is taken. To manage the catheter up to the heart, a long exposure time (several minutes of fluoroscopy) is needed. If the contrast agent were injected into a peripheral vein, the procedure would be less invasive with a sensible reduction of risk, but the contrast medium would be much more diluted (around 10 mg/ml [5]) and hence the imaging would require higher exposures.

Monochromatic X-ray beams can be used to implement dual-energy techniques to increase the image contrast. In this way, the contrast medium concentration and the dose delivered to the patient can be reduced. The K-edge subtraction angiography (see Ref. [5] and references therein) consists of two mono-energetic photon beams, one immediately below, the other immediately above the iodine K-edge at 33.17 keV. Two images are taken simultaneously with digital detectors and then a digital (logarithmic) subtraction is performed pixel-by-pixel, thus extracting the iodine signal. The application of K-edge subtraction angiography with intravenous injection of the iodate solution has provided quality images using synchrotron radiation with a considerable reduction of contrast medium concentration and exposure, i.e. a reduction of the involved risk.

As a test object we used a plexiglass step wedge phantom (see Fig. 7), 40 mm high and 30 mm wide and with a thickness ranging from 10 to 45 mm. The three cylindrical cavities, having a 1 mm diameter, have been filled with iodate contrast medium (with a concentration of 370 mg/ml) to simulate patient vessels.

Profile images of the test object have been taken with a 31 keV (below the iodine K-edge) and subsequently with a 35 keV (above the iodine K-edge) quasi-monochromatic X-ray beam. The two-dimensional image is obtained by moving the test object along the orthogonal direction (see Fig. 7). The measured number of counts have been corrected for the X-ray beam intensity profile and for the different flux and detector efficiency at the two energies.

Resulting images are shown in Figs. 8 and 9, together with their mean vertical profiles. Only two cavities are visible in the image because the third one is out of the geometrical acceptance of the detector. It can be seen that in the image at 35 keV the measured detail contrast (i.e. the difference between the minimum count level in the iodine filled cavity and the mean count level measured on the background) is about 2.5 times higher than the one measured at 31 keV. Furthermore, in the images is also present a vertical gradient due to the attenuation characteristics of the step-wedge phantom.

The image obtained by performing a logarithmic subtraction of the images collected at 31 and 35 keV is shown in Fig. 10 together with its mean vertical profile. It can be observed that no gradient

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5 This test mainly is intended to evaluate the feasibility of the imaging and the performance of the detector in a preliminary experimental setup. The evaluation of quantities which are important for the medical application (e.g. dose, scanning time etc.) is beyond the goals of this first test.
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 effects, the detail visibility is dramatically improved in the final image with respect to the primary images.

8. Conclusions

A detecting system for position-sensitive X-ray measurements based on a 132-strip silicon detector and two RX64 readout chips has been developed and tested. The noise and gain characteristics of the readout chip have been measured with the internal calibration of the RX64 obtaining 61.6 μV/el of gain and 131 electrons of equivalent noise charge.

A more precise evaluation of the gain has been extracted from measurements with fluorescence X-ray sources, obtaining 17.0 mV/keV, which when converted into μV/el units turns out to be in good agreement with the one obtained with the internal calibration. The noise counts are completely discriminated by an 80 mV threshold, corresponding to an energy of ≈4.7 keV.

Measurements with quasi-monochromatic X-ray beams in the 18–36 keV energy range have been taken with strips being either orthogonal or parallel to the incoming photons. The X-ray energy is in excellent agreement with the expectations based on the results of the calibration with the fluorescence sources and the energy spread is small with respect to the intrinsic noise of the detecting system.

Preliminary iodine K-edge subtraction imaging results obtained with a plexiglass phantom with tiny cylindrical cavities filled with iodate solution have also been presented. They show that the use of this technique provides an enhancement of the image contrast by canceling the background structure and isolating the iodine signal. This technique could lead to a reduction of contrast medium concentration and dose delivered to the patient and allow to perform angiography with intravenous injection of the iodate solution. The spatial resolution of the detecting system and other effects (such as charge sharing and Compton scattering) which can deteriorate the quality of the image are presently under study.

Work is in progress to develop a larger detecting system (4 cm large detectors with 400 strips) and to build a large field dichromatic source, with size and fluxes conform to the clinical application.
References