CMOS X-ray Image Sensor Array

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***Abstract*—This paper describes a pixel array for x-rays imaging consisting in an 8x8 photodiode array fabricated in CMOS technology, with the respective readout circuit. Above the photodiodes, an array of scintillating CsI:Tl crystals are placed. So, the x-ray energy is first converted to visible light by the scintillating crystals, which is then detected by the photodiodes. The photocurrent produced by each photodiode is amplified, converted to a voltage and stored in a capacitor. Finally, the voltage of each capacitor is read by the readout circuit.**

I. INTRODUCTION

One of the first x-ray sensors developed was based on a silicon Charge Coupled Device (CCD) imager [1]. The silicon has a low x-ray absorption coefficient, but for each 1 MeV of x-ray photons absorbed, about 277 000 electrons are excited [2]. This enables the construction of x-rays sensors with better sensibility than the traditional radiographic silver-halide films. However, the small number of detected photons in the imager results in a significant quantum noise [2]. In order to reduce the quantum noise, the radiation dose can be increased or the quantum efficiency of the sensor can be improved. The increase in the x-ray dose is obviously not desired for medical applications. The quantum efficiency of the sensor can be increased by adding a scintillating layer above the imager. A scintillating layer is a layer composed by a material that produces visible light when the x-rays hit it. An example of scintillator material is the thallium doped cesium iodide (CsI:Tl), which produces green light in the presence of x-rays or other kind of ionizing radiation. Since the x-rays are first absorbed by the scintillating layer, which has a high absorption coefficient, and then converted into visible light, the quantum efficiency of the detector is improved, once each x-ray photon produces several thousands of visible light photons. A drawback of this approach is that the spatial resolution of the device is approximately equal to the thickness of the scintillating layer [3] (Fig. 1), i. e., the thicker the scintillating layer, which absorbs more x-rays, the poorer the spatial resolution. Several techniques were tested in order to increase the scintillating layer thickness without decreasing the spatial resolution [4, 5].

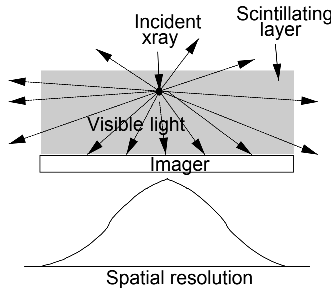

Fig. 1. X-ray imager with a scintillating layer and its spatial resolution.

Fig. 2 shows a schematic representation of the proposed detector structure. The photodetectors consist on a CMOS photodiode array. The scintillators are placed above the photodetectors. In this case, the chosen scintillator was the CsI:Tl, due to its high light yield and relatively high density and atomic number of its elements, which is necessary to absorb the x-rays.

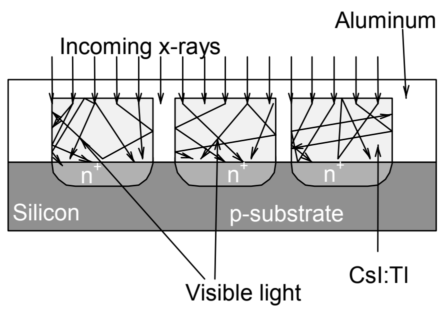

Fig. 2: Schematic representation of the detector structure.

A reflective material is then used to coat the scintillator. It works like a light guide avoiding the visible light dispersion and the interference between each neighbor pixel. Moreover, it improves the spatial resolution as well as it increases the intensity of the transmitted light to the photodetectors. In addition, the quantity of the incoming x-ray radiation can be reduced, while keeping the same sensitivity of the photodetectors signal readout. This material must have a low density and a low atomic number to allow the penetration of the x-rays. The aluminum is a good

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candidate. In this case, the x-rays cross the reflective material placed on the top and reach the scintillators, where they are absorbed. For each x ray absorbed photon, many visible light photons are produced, traveling in all directions. Some of them arrive directly at the photodetector, while others reach the reflector. After some reflections, disregarding the losses in the reflection, almost all the visible light photons reach the photodetector. A technique used to fabricate these scintillating light guides was recently patented [6].

There are several reasons to substitute the CCDs by CMOS photodetector arrays. Some of these reasons are described in the next paragraphs. CCD technology is robust for imaging applications and offers several charge-transfer architectures and specialized structural layouts. Usually, it is optimized for photodetection and features very low noise and a very high fill factor (the ratio of the photosensitive area to the total array area). CCD manufacturing outfits a number of technological issues in order to achieve high charge-transfer efficiency and to minimize crosstalk and noise. These issues increase the CCD fabrication complexity and cost. Furthermore, because of serial nature of the charge-transfer process, a conventional CCD does not feature random access to each pixel [1]. Typically, functional circuitry cannot be integrated reliably on-chip and when it is, the performance is poor due to the difficulty of driving the large capacitive loads of the CCD, unless a combined CCD/CMOS process is implemented, which clearly introduces relevant extra costs.

Applications that require imaging sensors would benefit if all the readout electronics could be integrated with the image sensors in a single-chip. Advantages of such integration are lower power consumption, cost reduction, reliability improvement and data conversion speed up. These requirements can be achieved by using a CMOS process. Besides being widely available, a standard CMOS process is generic, supports several photosensitive structures, favors the integration of multiple electronics functions with a high yield and it is, currently, the cheapest of the competing technologies. If a standard technology for the fabrication of image sensor system has to be followed, CMOS is a more appropriate option due to its wide availability, reduced component and packaging costs and, more important, its capacity to accommodate multi-functional circuitry on-chip. Therefore, it stands as a reasonable choice for the fabrication of the proposed integrated image sensor system. However, standard CMOS is optimized for electronics and not for imaging. Consequently, it has a tendency to higher noise levels than CCD technology. Recent progress in on-chip signal processing allows the correction of fixed-pattern noise with negligible system impact, which has led to a reduction of CMOS image sensor fixed-pattern noise to acceptable levels.

The recent development in CMOS image detectors opens a new way to construct digital x-rays imagers. The replacement of CCDs with CMOS detectors in x-ray imagers is desirable for several reasons:

- The operating power is 5 to 10 times lower than the CCD (with processing electronics).
- The CMOS is a standard fabrication process, while CCD requires special manufacturing.
- CMOS fabrication costs are 5 to 10 times lower.
- It is possible to integrate analog and digital processing electronics in CMOS.
- It is advantageous in terms of radiation dose reduction if the whole radiography is taken in a single shot. In opposite to CMOS, this is not possible with CCD, once it reads the image column by column.

The drawback is the difficult to match the high performance characteristics of CCD in terms of image quality [7].

II. Sensor Description

The sensor consists in an array of 8x8 pixel blocks, containing each of them a photodiode and a pixel readout circuit. The following subsections will describe each of the sensor modules.

A. Main module

The circuit consists in a 8x8 pixel matrix and a 6 bit counter, which counts from 0 to 63, in order to address each pixel of the matrix (Fig. 3).

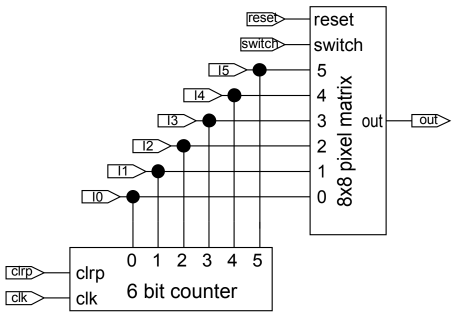

Fig. 3: Main circuit module.

There are two working possibilities for this module: with the input *clrp* at the high level, the counter runs and each pixel of the matrix is addressed sequentially. With the *clrp* input at the low level, the counter stops and its outputs go to a high impedance mode. In this case, it is possible to address each pixel individually by selecting its address in the inputs I_o to I_5. Notice that in this case, the matrix row is selected by the inputs I_o to I_2 and the matrix column is selected by I_3 to I_5. The readout speed is determined by the clock frequency, which can be in the order of few megahertz, allowing the readout of the whole matrix in few tens of microseconds.

The *Reset* and *Switch* inputs control the current to voltage converter at each pixel in order to hold the voltage value of each photodiode until the whole matrix is read. Contrarily to a visible light imaging detector, in which the light intensity at each pixel can be read sequentially by lines and columns, in a x-ray detector all the pixels must store their values at the same instant time. This time must be synchronized with the x-ray source.

B. 6 bit counter

Fig. 4 shows the schematic diagram of the 6 bit counter. It consists in a 6 JK flip-flops operating in a synchronous configuration.

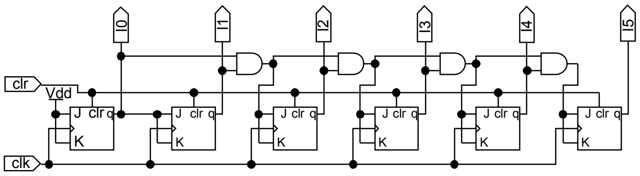

Fig.4: 6 bit counter schematics.

When the *clr* input goes to the high logic level, all the outputs of the flip-flops stay in a high impedance level.

C. Photodetector matrix

Fig. 5 shows a part of the 8x8 photodetector matrix.

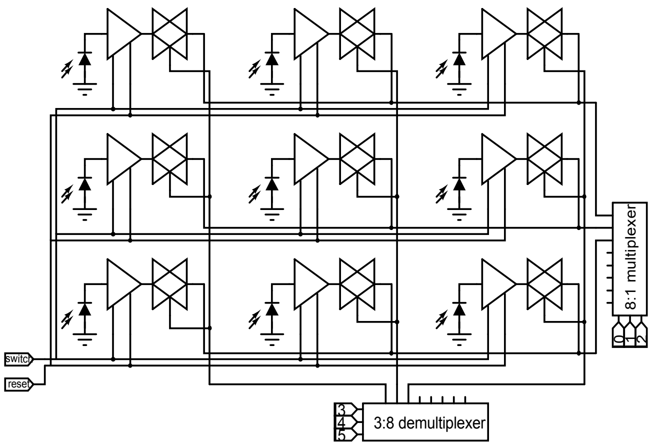

Fig.5: Part of the 8x8 matrix.

For a proper working, it is necessary a 3:8 demultiplexer and an 8:1 analog multiplexer. The 3:8 demultiplexer receives at its inputs the outputs I_3 to I_5 of the counter and selects the corresponding column of the matrix by placing a logic high level at all the transmission gates of that column. The 8:1 analog multiplexer receives as addressing lines the outputs I_o to I_2 of the counter and according to its logic information connects the corresponding line to the matrix output.

D. Circuit for each pixel

Fig. 6 shows the circuit of each pixel. It is composed by a photodiode, a transimpedance amplifier and a transmission gate.

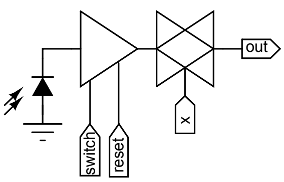

Fig. 6: Circuit for each pixel.

When the visible light produced by the scintillator reaches the photodiode, it produces an electric current, which is converted to a voltage by the transimpedance amplifier. With the help of the *Reset* and *Switch* inputs, the voltage value is maintained constant until the entire matrix is read. The *x* input is connected to one of the demultiplexer outputs. When this input is at the high level, the transmission gate connects the voltage at the output of the transimpedance amplifier to the corresponding analog multiplexer input.

E. 8x1 analog multiplexer

Fig.7 shows the 8:1 analog multiplexer circuit.

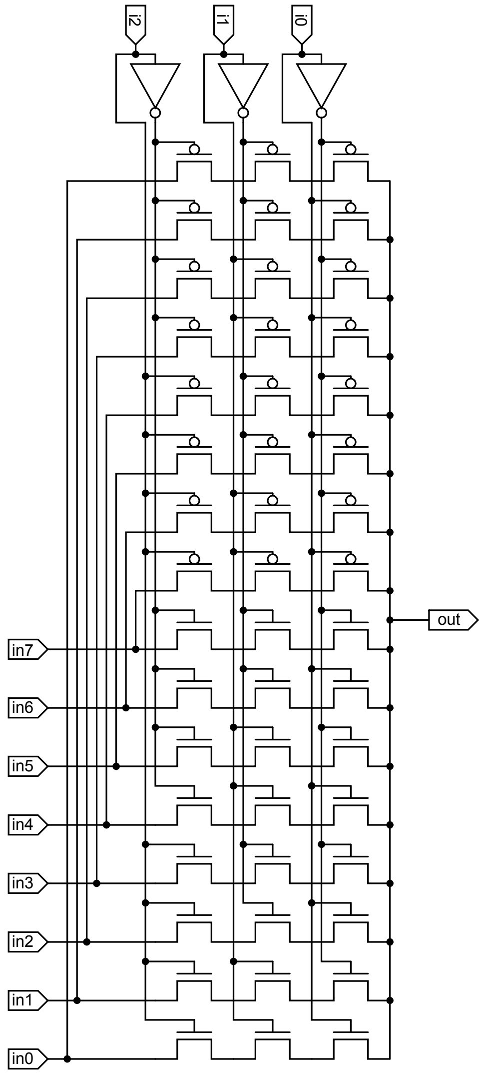

Fig. 7: 8:1 analog multiplexer circuit.

The multiplexer, according to the logic levels at the lines I_o to I_2, will connect the input n_o to n_7 to the output.

F. Transmission gate circuit

Fig. 8 shows the schematic diagram of the transmission gate circuit.

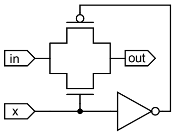

Fig. 8: Transmission gate circuit.

As it can be seen, when the x input is at the high logic level, both the transistors conduct and the input is connected to the output. By the other hand, when x is at the low logic level, both transistors are off and the output of the transmission gate is at a high impedance state.

G. Transimpedance amplifier

Fig. 9 shows the schematics of the transimpedance amplifier circuit.

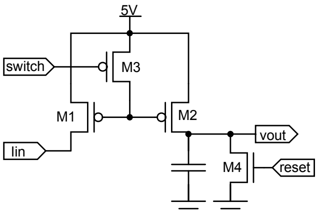

Fig. 9: Transimpedance amplifier circuit.

The photodiode is connected to the current mirror input *Iin*. This current is amplified by the current mirror comprised by M_1 and M_2 and charge the capacitor. Considering a constant current I in a time interval t, the voltage at the capacitor terminals will be $V{=}I\,t$. In this way, the photodiode current is converted to a voltage. The transistor M_3 controls the current mirror. When the input *Switch* is at the low logic level, M_3 will conduct and the voltage at the gates of M_1 and M_2 will be 5V. In this case, both the transistors are off and the current at the capacitor will be zero. Therefore, the circuit is in the hold state. When the *Reset* input is at the high logic level, M_4 will conduct and the capacitor is discharged.

III. Results

This section shows the waveforms at the output of the transimpedance amplifier of one pixel and at the output of the 8:1 analog multiplexer.

A. Transimpedance amplifier

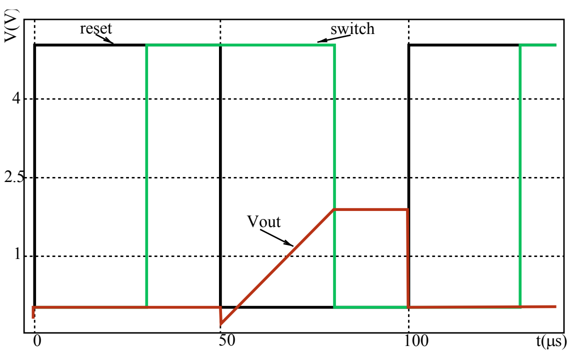

Fig. 10: Transimpedance amplifier output waveform.

Fig. 10 shows the waveform at the output of the transimpedance amplifier. When the *reset* signal goes to the high level, the output is zero. When the *switch* signal goes to the low level, the output voltage is hold by the circuit.

B. Output of the main module

Fig. 11 shows the waveform at the 8:1 analog multiplexer output when the first 16 pixels are read. The current in each photodiode has a value in the range from 1 µA to 75 µA.

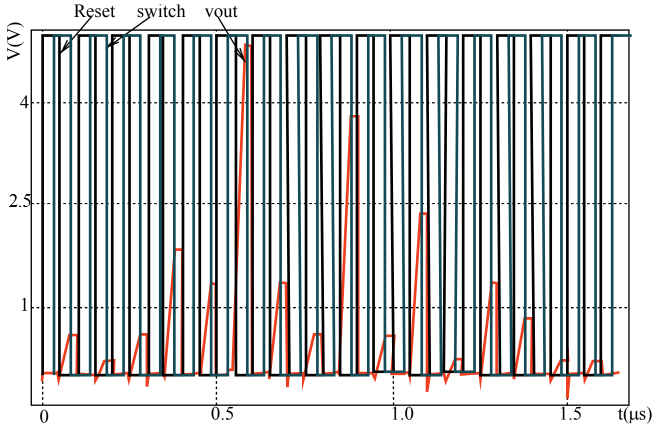

Fig. 11: 8:1 analog multiplexer output waveform for the first 16 pixels.

IV. Conclusion

The main difference in the readout circuit of a visible light imager and an x-ray imager is that in the first one it is possible to read the pixels sequentially and in the second one it is not possible. In an x-ray imager, all the pixels of the matrix must be read at the same time, synchronized with the x-ray tube that produces the x-rays. By the other hand, the image storage must be as quick as possible, once the radiation is harmful to the patient. In this work, we reported a circuit that amplifies and holds the entire pixel matrix values at the same time, in order to be possible to later read them in a sequential way.

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