Color X-ray photon counting image sensing

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Introduction

Present state of the art medical X-ray image sensors are “charge integration” based, which results in an overall noise that is composed of the quantum limited photon shot noise, electronic read noise and excess noise due to the non-reproducible charge packet sizes per absorbed X-ray. Ultimate sensitivity and signal to noise ratio is obtained when each incoming photon is counted, the so-called quantum limit.
Another advantage of photon counting is that one can classify the incoming photons according to their energy, and in that way realize spectral sensitivity or “color X-ray” [5] without having a penalty in dose or multiple exposures.

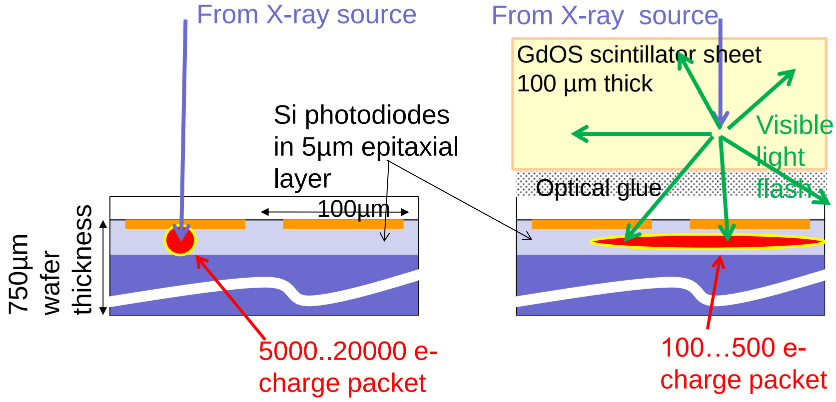


Figure 1 Direct (left) vs indirect (right) X-photon detection in the present device.

State of the art X-photon counting imagers [1, 2, 3] use *direct detection*, i.e. the X-ray to charge conversion happens by the photo-electric effect in a high-Z semiconductor photo conductor or photodiode. For medical X-ray energies, photons generate charge packets in the order of 5000 to 50000 electrons, which are easily processed by hybrid connected electronic readout circuits. In indirect detectors, or scintillators, the X-photon is absorbed in the scintillator volume, where it creates a secondary visible light flash, which is detected by a Silicon photodiode in an underlying readout IC (Figure 1). Scintillators are more economical to manufacture than high-Z detector hybrids, yet the efficiency of the overall indirect detection process is poor: charge packets per X-photon are in the range 100 to at most 1000 electrons, depending on scintillator characteristics [4, 5]. Photon counting with scintillators will thus require a very low noise and accurate threshold detection concept, with an accuracy and noise floor far below 100 electrons [6, 8].

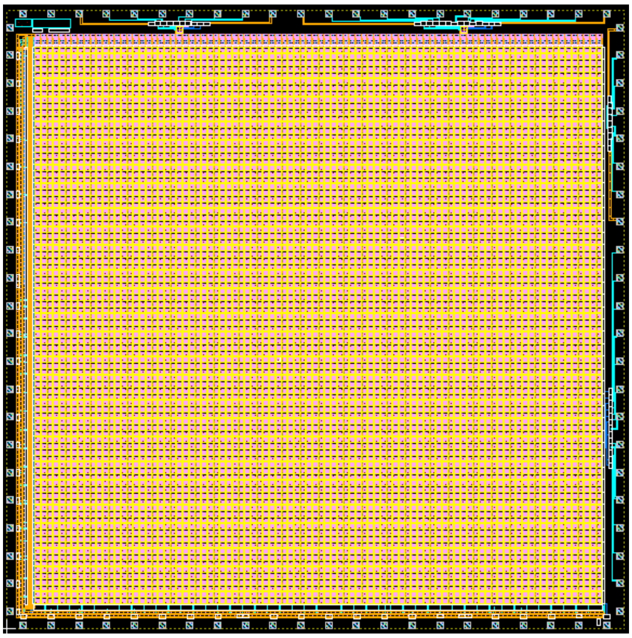
Challenges addressed

The realized QX2010 device demonstrates solutions for following challenges:

→ How to realize compact monolithic photon counting pixels with decent fill factor for use with a scintillator. How to make the counting circuitry small enough by employing non-linear analog domain counting (Figure 6).

→ To obtain a high charge sensitivity and comparator accuracy with an equivalent noise floor far below 100 e^-_{RMS}..

→ To realize charge package size discrimination and thus realize “color x-ray” capability. In the present circuit we implemented two charge magnitude threshold levels (Figure 3)



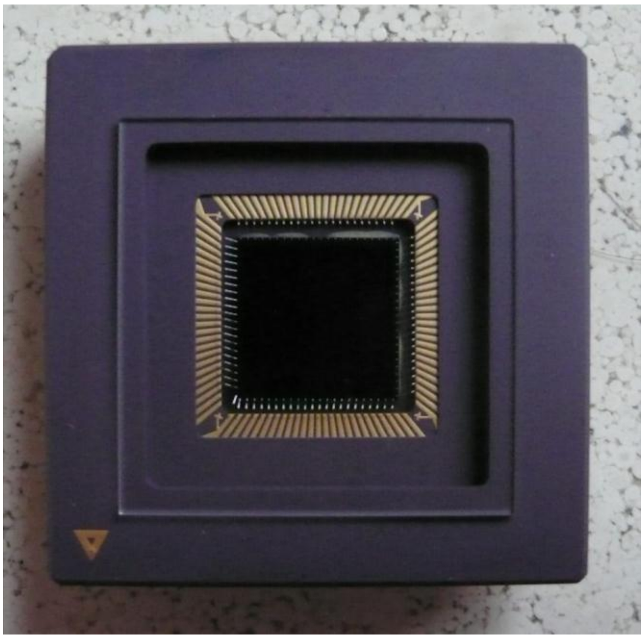


Figure 2 Layout drawing and picture of the QX2010 92x90 pixel color x-ray photon counting demonstrator.

The QX2010 (Figure 2) is a 1cm^2 large demonstrator array. It is processed in a standard CMOS technology with high resistivity epi. The QX2010 has 90x92 pixels on 100µm pitch Pixels have a fill factor of ~75%. The pixel topology is shown in Figure 3. Each pixels contains 1 photodiode + pulse shaper, yet two parallel channels with comparators and analog counters. The multiplexer output either the two analog counts, or the real time signals in the comparators, for debugging and analysis purposes. The total number of transistors in this pixel is 45. The pixel layout is shown in Figure 4.

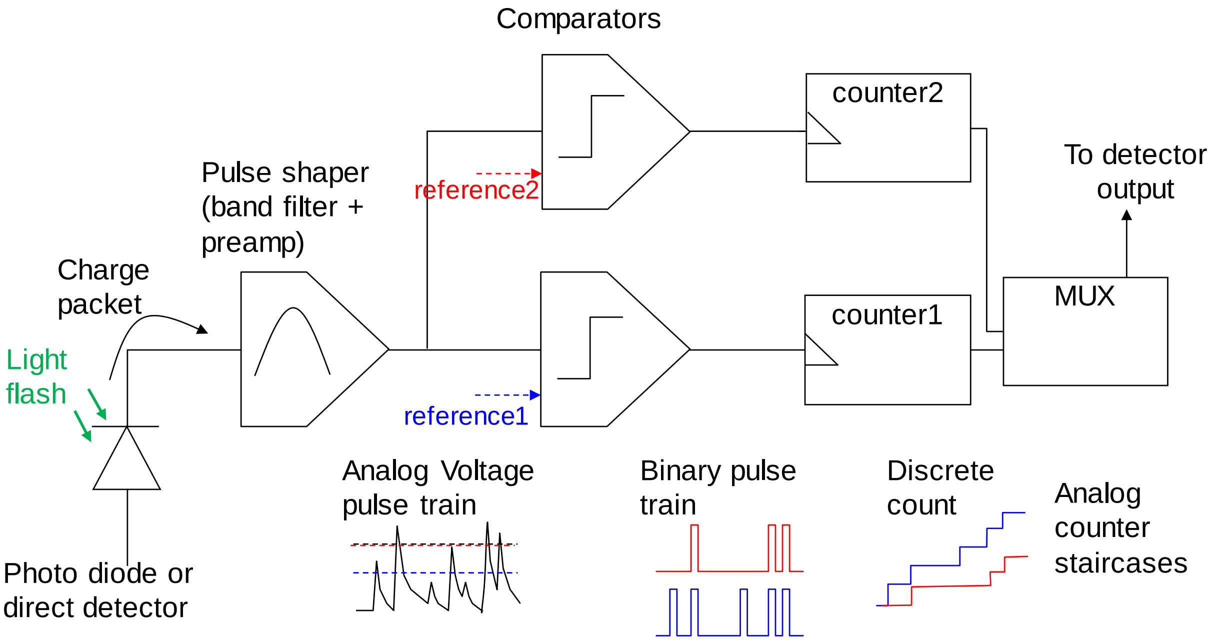


Figure 3 The 2-Color X-photon counting pixel circuit topology

Key circuit parts determining the performance of this pixel are the pulse shaper (Figure 5), that must be at the same time very compact in layout, and highly sensitivity to discriminate charge packets of ~100 electrons, and the compact analog counter, that creates a stepped staircase voltage (Figure 6).

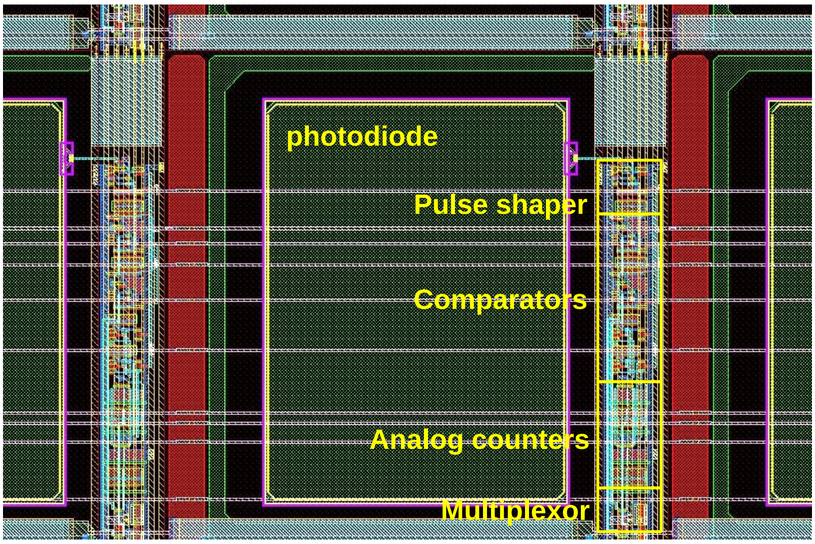


Figure 4 Layout view of the QX2010 pixel.

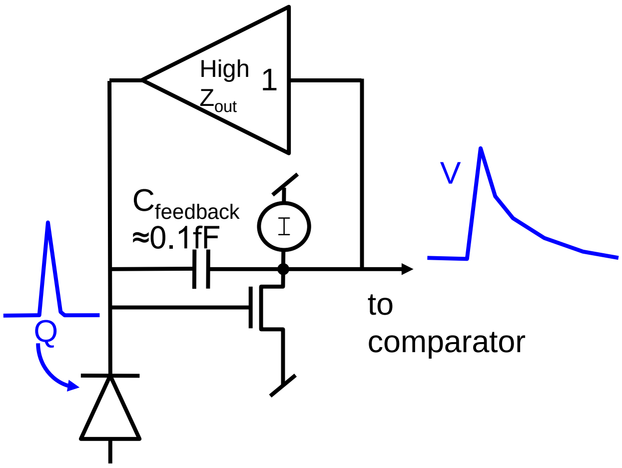


Figure 5 Pulse shaper circuit topology

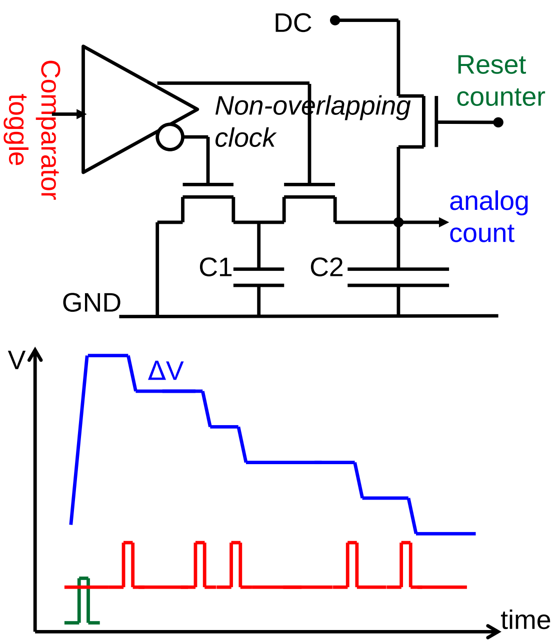


Figure 6 Analog counter and its key signal traces.

Measurement results

Functional measurements are done in three ways.

1. Using an electrical input. Pixels at the edge of the matrix have a precise capacitive charge input. This allows to check functionality without optical input, and allows to accurate calibrate the charge packet to voltage pulse height conversion factor. Using the electrical input one estimates the equivalent pulse discrimination noise floor of the system at about $15e^{-}_{RMS}$. The electrical input allows running the circuit at its highest possible speed, which is faster than 1MHz sustained count rate, with less than 1 to 10µW/pixel.

2. Using an optical input. Here we use pulsed LED illumination (Figure 7). LEDs are generally fast enough to generate light pulses shorter than 1µs. As one pixel in the array is deliberately covered with a metal shield, we can thus evaluate the tolerance for crosstalk of one silent pixel in a full array of counting pixels. The crosstalk measured as such is very low.

3. Using X-ray input with dose and energy representative for medical applications. The bare array is highly responsive to direct detection events, though the probability for direct absorption is very low. A counting staircase is shown in Figure 8. The more relevant setup is as an indirect detector, where we demonstrated photon counting through a simple GOS scintillator sheet [8].

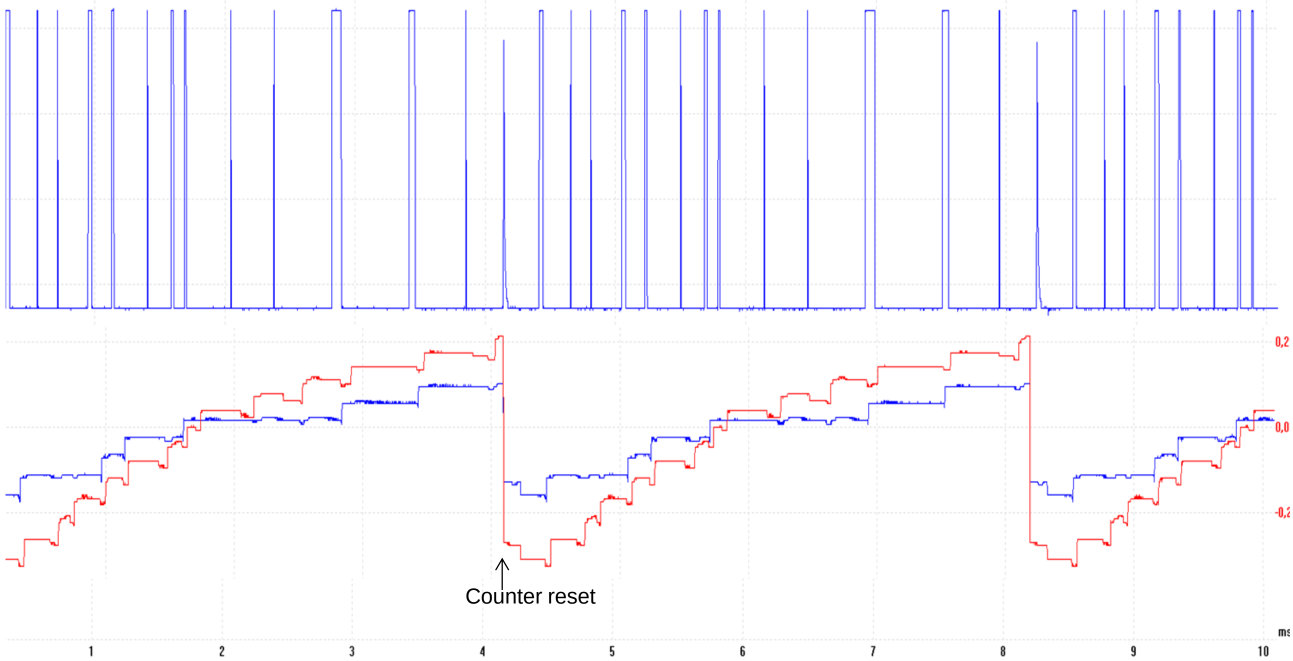


Figure 7 Real time observation of counting staircase for light pulses pattern generated with a LED. Upper trace: LED pulse pattern, where a wide pulse generates a larger charge packet. Lower traces red/blue: counter staircases for the two channels (trace acquired AC-coupled). The red channel counts every LED pulse. The blue channel counts only the wider LED pulses.

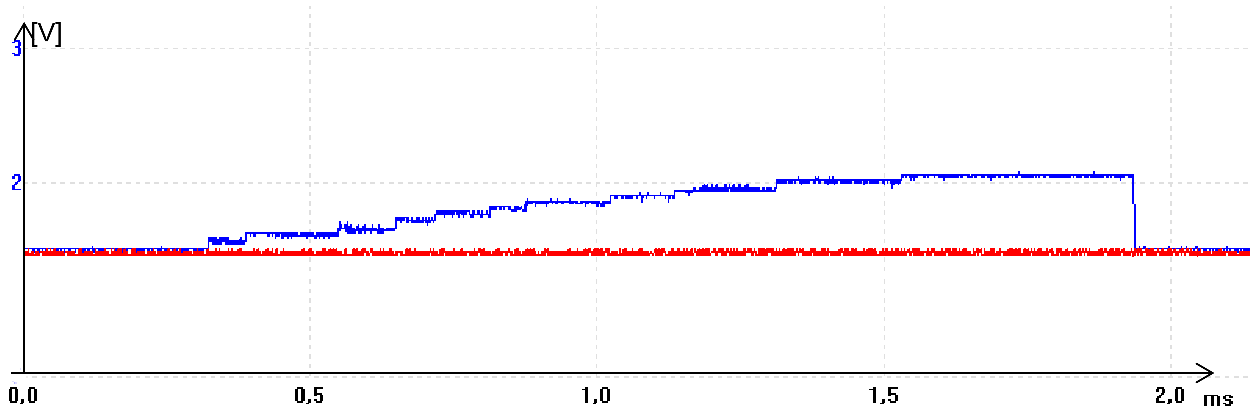


Figure 8 Real time trace of the counter staircase of a QX2010 pixel as direct detector (i.e. without scintillator). X-ray source at 1.5m distance; 100kVp, 70mA.

References

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