Photon counting and color X-ray imaging in standard CIS technology
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Abstract
With the extensive usage of radiography in many medical diagnosis protocols, the urge to achieve the lowest possible dose imaging is a driving force for the X-ray image sensor community. The ultimate signal to noise ratio that one can theoretically achieve is the quantum limit, where each and every photon reaching the imager is counted. Moreover, photon counting comes with an additional benefit: the possibility to sort incoming X-ray photons based on their energy, achieving "color X-ray" imaging, a real added value for the diagnostician.

This is a compilation of previous publications by Caeleste and public information.

Why color X-ray imaging?

The clinician’s point of view
The present state of the art digital radiography (DR) is made of low noise charge integration pixels, either made of monocrystalline (CCD, CMOS) or amorphous (TFT) Silicon, combined with scintillation or direct detection material. With the strong focus on dose reduction, the main R&D area has been the lowering of the noise floor. The “charge integration” images are radiation intensity coded, and bear no energy information. Visualization equipment can of course render “false colors”, but this does not have the added diagnostic value of true color X-ray.

True color X-ray imaging is an advantage for the clinician. Applications range from medical diagnostics, bone densitometry, mammography. We recently participated in a study, led by the Departments of Anatomopathology and of Radiology of the Universitair Ziekenhuis Brussels (UZB), showing that Multi-energy X-ray has an enhanced visibility of tumor tissues in mammography [1][2][3]. Double blind tests have demonstrated that in 23% of the cases, the traditional X-ray picture underestimate the stage of the tumor compared to the color X-ray with energy discrimination. Color X-ray also revealed multifocal lesions that were missed by the classical method [3].

The physicist’s point of view
Essentially color X-ray images have the capability to sort tissue or material based on their chemical atomic composition. As X-ray spectral absorption depends largely on the element’s (atom’s) Z-number, one can by examination of the absorbed of transmitted spectrum deduce information on the relative concentration of the elements. This is explained in Figure 1 where the spectral absorption of the most prevalent elements in biological tissue is shown. In biological tissue one major type of information is the ratio of Carbon over Oxygen. As an example, a comparison of mammary resection specimen, in X-ray black & white (B&W) and color images are shown in Figure 2 and Figure 3.

Today’s X-ray equipment allows dual and multiple energy X-ray imaging by dual or multiple exposure with different energy spectra. This is an acceptable technique on operation specimen, but widespread use in medical imaging is precluded by the additional radiation dose due to the multiple exposure. The ideal way to obtain multi-energy information in the X-ray image is by energy-classifying the photons as they arrive, within the same single and original exposure.
Figure 1: Left: comparative spectral absorption of X-rays in the most relevant elements in biological tissue. Right: Relative composition (weight%) of elements in common biological tissues.

Figure 2: B&W X-ray and color X-ray images from the same mammary resection specimen. In this color image, hue codes the oxygen/carbon ratio (here blue=carbon-rich, red=oxygen-rich).

Figure 3: Breast carcinoma specimen “Poorly differentiated invasive duct” type [2]. B&W X-ray and color X-ray. Cyan = C-rich, Red = O-rich.
Detecting X-rays

Detection methods

Various methods exist to detect X-ray with Solid state imagers. Two major families can be distinguished: devices based on direct detection and devices based on scintillation. In direct detectors, the readout circuitry is electrically coupled to a “heavy” (high atomic number) semiconductor, such as Cadmium-Telluride (CdTe) which exhibits a good absorption of high energy photons. The X-photon to charge conversion for such material is large as shown on Figure 5.

In direct detection material, an X-ray photon with a typical energy for medical imaging produces about 5000 to 20000 electrons. Direct detection materials are generally expensive, requiring bump bonding or dedicated deposition techniques. Scaling to large formats may be a problem. Monocrystalline CdTe detectors can be found up to a few cm$^2$ in size, which makes the creation of a large format detector prohibitively expensive.

The alternative is to use scintillation, also presented in Figure 4. Scintillators are far more economical to manufacture than high-Z direct 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]. Moreover, scintillators have an inherently worse MTF than direct detection material, due to optical crosstalk within the scintillation material. The crosstalk can be lowered using CsI needle-like crystal growth, or using a dedicated optical confinement approaches such as [5].

Figure 5: optimal electron packet size for direct and indirect detection, for various and most popular materials, and as function of the photon energy.
Classic X-ray pixel circuits

Classic “charge integration” based digital X-ray modules have been deployed in volume in medical imaging since about 15 years with the development of large scale, affordable TFT plates. The use of CMOS/CCD sensors is more recent, and in the first place in the field of dental X-ray [7] where the size of the sensor is compatible with typical CMOS manufacturing reticle sizes. For cost considerations the vast majority of detectors used in commercial X-ray systems is based on scintillation. Apart from their size, charge integration X-ray pixels differ little from visible light imaging pixels. One benefits thus directly from the innovations in visible light imaging, such as pinned photodiodes, SoC, readout structures, advances in scaling and yield. The following picture is a typical CMOS (X-ray) image sensor pixel by Caeleste.

![Image](image_url)

**Figure 6:** pinned diode pixel, principle (left), layout view (right)

Photon counting X-ray pixel circuits

Photon counting pixels fundamentally differ from classic integrating pixels, as they result from a different trade-off space. Classic integrating pixels must handle the total amount of charges generated by the indirect visible photons hitting the pixel. As the result, their typical full-well charge may be in the order of 1Me+ or more, in thorax or mammography X-ray systems. In photon counting pixels, each charge packet emanating from one X-photon is sampled, measured, and accumulated as “one” count. One consequence is that the charge sensing capacitance must be set as low as possible, as it must only sense one such packet at a time. The following figure is the typical photon counting pixel schematic concept [7].

![Image](image_url)

**Figure 7:** typical photon-counting pixel block diagram

A pulse shaper converts the photocurrent pulse or charge packet to a voltage pulse. Such pulse shaper is essentially a band-filtered trans-impedance amplifier. A reference voltage common to all pixels sets a comparator threshold above which the charge packet and therefore the original X-ray photon is seen as an “event” and the counter increments. The final counter value is readout over a multiplexer.
It is possible to extend the scheme of Figure 7 to create dual energy pixels:

![Dual energy pixels diagram](image)

**Figure 8: dual color photon counting pixel block diagram [8]**

By choosing different reference voltages charge packets can be distinguished based on their respective intensity. One channel counts all the incoming photons within the bandwidth limitation of the system. A second channel counts only the high energy photons, since the charge packet size depends on photon energy as per Figure 5.

**Scaling photon counting pixels to a large array**

Photon counting detectors and cameras exist since a long time. One of the pioneers is the Medipix [9] project, started at the CERN to detect gamma photons and other high energy particles. Other groups have pioneered hybridization with heavy direct detectors as CdTe [10].

Caeleste’s first steps in hybrid photon counting were not in the domain of X-ray but in visible light avalanche photo diode (APD) readout. Our first photon counting pixel was a highly integrated pixel with 479 transistors per 100µm x 100µm pixel, bump or wire bonded to a separate, thus hybrid, ADP chip. It was designed for and operated as a time of flight (TOF) application [11].

![Hybrid pixel images](image)

**Figure 9: Caeleste+SensL hybrid APD counting pixel array [11], picture of hybrid assembly (Left). Layout view of the 100um x 100 um pixel with 480 transistor per pixel (right)**

Theoretically one can pursue almost the identical concept to realize X-ray photon counting using hybrid direct detectors. Although such hybrid solution offers the best possible level of performance, such a complex structure will exhibit great challenges in terms of scalability and cost. To create one mammographic detector or chest X-ray detector, several 8” or 12”-waferscale detectors have to be assembled to form an array. Then,
with transistor counts greater than 400 transistors per pixel, the manufacturing yield of a perfectly functional device with those dimensions is more than likely uncompetitive. Moreover, the heavy semiconductor monocrystalline substrates are only grown in relatively small bars, allowing at best the manufacture of a few cm² large dies. At least 20…30 of such dies have to be assembled on one 8”-waferscale device. The price of such material is prohibitive for large arrays to compete with state of the art detectors, not mentioning the associated assembly cost and yield. A more promising route is the use of amorphous Se, which can be deposited as a layer, even on wafer scale. However, Se being not a real heavy element, the time constants associated with the amorphous state, and the instability of the deposited material poses also challenges.

**Photon counting with scintillators**

Concept

To overcome the scalability issue, Caeleste has created a Color X-ray photon counting technology aimed at the highest possible scalability and manufacturability with the following constraints in mind:

- No heavy semiconductor hybridization, the detector must work with optically coupled scintillators.
- Use a standard, CMOS Image Sensor process technology
- Front side illumination
- High yield is ensured by a low transistor count per pixel and fault tolerant design
- High scalability as obtained by stitching and minimal gap abutting
- Low power

Up to today Caeleste has designed and evaluated two generations of such detectors, starting from a 16x16 B&W photon-counting pixel array to the 90x92 pixel dual color 1cm² prototype described in this white paper. The actual floor plan of the “QX2010” device is presented in the following picture. The largest part is occupied by the photon-counting array. Top and right edges contain Caeleste’s process control modules and other test structures.

The device was realized in a standard CIS 0.18um and packaged in either a standard PGA or chip-on-board.
The chip-on-board package allows testing various scintillation materials. Tests have been performed using the following scintillation foils and sheets, optically glued directly on the CMOS detector:

- GdOS scintillator
- Cesium-iodide scintillator

Photon-counting does not require a fiber optic faceplate, as a directly detected photon will be counted as one single event whereas integration sensors are overwhelmed by the large amount of electron generated by a directly detected X-ray photon. The prototype datasheet can be found in annex 1.

**Design details**

One pixel layout [9] is presented on the following figure. This dual color X-ray pixel contains 45 transistors. The earlier black and white version had 27 transistors only. The electronics are remarkably compact. The metal limited fill factor of this 100 µm x 100 µm pixel is 80%, enabling the sensor to operate optimally in front side illumination despite the transistor overhead footprint. Moreover, the design is very scalable, both to smaller pixels sizes and very large arrays.

Such compactness is achieved thanks to Caeleste's know-how and experience, enabling unique innovative solutions that have led to several patent applications. Although the generic architecture presented in Figure 8 rules the design of the photon counting pixel, several blocks have been reduced dramatically in transistor count. A key innovation is the use of non-linear analog counters instead of the traditional digital counters. The following picture shows the concept used in the Caeleste pixel. The digitized pulse from the comparator triggers an ultra-compact non-overlapping clock generator that retrieves a charge packet from a larger capacitor. The result is a stair-case signal, where each quantized step represents one photon.
A very interesting feature of this method is the fact that the response is non-linear. Indeed, as the number of count increases, the size of the charge packet retrieved from the capacitor becomes smaller and smaller, thus exploiting the increase of X-photon shot noise. This is useful since it increases the maximum count for an identical capacitor size and read noise, compared to a linear implementation. The non-linear output can be linearized at system level. Of course, the usage of an analog counter instead of a full-blown digital implementation raises the question of noise immunity of the readout method. Indeed, steps remain quantizable or below X-photon shot noise as long as the difference between two consecutive steps remains above the noise floor.

The following trace shows a measurement of pixel output as a function of time, whereby the X-photons are emulated by a LED pulse train, shown he non-linear analog count fashion:

**Figure 14:** counter output at 300kHz LED pulse rate during initial chip characterization

**Electrical characterization & visible light testing**

A homogenous source of light was created by using a front face diffuser. The light is a pulsed LED, which allows the modulation of charge packet size with the duration of the light pulse. A logic pattern generator is used to generate these LED pulses. The following trace was acquired using visible light as an input. It shows that two thresholds can be used to discriminate packet sizes inside the same pixel.

**Figure 15:** Caeleste photon-counting pixel output with pulsed visible light input
The top traces are the analog “staircase” counter outputs. The red staircase counts all the pulses whereas the blue staircase only counts only the larger events. The corresponding comparator internal signals are shown at the bottom. Pixel to pixel offset on the output staircase is evaluated. This is due to the Vth variations of the PMOS source followers. Such FPN can be calibrated by “dark frame subtraction” or by CDS/DS as in classic image sensors. Alternatively some forms of on-chip FPN cancellation are considered too for future devices. The following table summarizes the FPN results for both channels.

<table>
<thead>
<tr>
<th>Channel A FPN</th>
<th>Channel B FPN</th>
</tr>
</thead>
<tbody>
<tr>
<td>23,753 mV RMS</td>
<td>28,361 mV RMS</td>
</tr>
</tbody>
</table>

The power consumption is critical when it comes to estimate what would be the consumption of a wafer scale device. In the actual setup the QX2010 consumes 7,319mW total power, or a power consumption per pixel of 0.88\(\mu\)W.

In order to calibrate the charge sensitivity and calibrate charge packet size we use an electrical input implemented as a test feature in the pixels at the right edge of the device. The capacitance has a design value of 0.1fF. A precise voltage step allows injecting an exactly known charge packet on the sense node.

![Figure 16: electrical input test setup](image)

From our tests, the minimum detectable packet size with 100% detection probability is 100 e-\(^-\). The minimum packet size difference that can be detected by the pixel, meaning the spectral separation of the pixel is 15 e-\(^-\). Another critical success factor is the ability to count photons at the highest possible rate. The following picture shows a count-rate of 300 kHz using. The comparator is responding accordingly. The analog counter is also functional as shown on Figure 14.

![Figure 17: 300 kHz electrical input internal comparator response](image)
Test under X-ray illumination

The devices were tested under X-ray in several conditions:

- No scintillator, direct detection only
- GdOS scintillator
- CsI scintillator

As stated earlier, one of the advantages of photon-counting X-ray detection is the fact that one can avoid the fiber optic face plate if desired. Indeed, direct detection only create a single hit on the photodiode, generated a single pulse instead of the large cloud of electron usually gathered by integration sensor. The following trace shows the response of the counter within the pixel to direct X-ray events:

![Figure 18: Photon counting pixel counter output, direct detection](image)

The indirect X-ray illumination measurements were mainly using GdOS scintillators, and later CsI. Our experiments in fact revealed the decays time of those scintillators. Indeed, photon-counting usage often raises the question about the detection speed and the risk of losing X-ray photons either by inappropriate CMOS circuit bandwidth or simply by the limitation of the scintillators. In the following plot were compared the comparator outputs for CSI and GdOS scintillator under the same X-ray illumination conditions.
The above plot shows that the GdOS scintillator is not fast enough. Although the pulse amplitude is consistent, the GdOS suffers from a long decay time impacting the maximum counting rate. CSI shows a significantly higher number of hits per pixel under the same illumination condition. With very low doses GdOS might still be useful. However, CSI scintillator seems much more suitable for higher flux applications. The conclusion is also that the electronic are fast enough. The scintillator is the bottle-neck when it comes to detecting X-ray photons.

The pixel is capable of discriminating X-ray photons based on their energy as shown on the following figure. The thresholds of the two channels are set separately. The blue channel only counts high energy photons, the red channel counts every photons. With our existing laboratory equipment, it is difficult to assess what the corresponding energy threshold is. However, our electrical tests using electrical input shows that the pixel is capable of discriminating two charge packets distant from 15 electrons, which represent a 2kVp difference according to Figure 5. This is more than sufficient to distinguish the interesting material from Figure 1.

Figure 19: photon counting pixel comparator output under X-ray illumination with GdOS scintillator (top) and CSI scintillator (bottom)
Conclusion

In this paper we presented the advantage of “Color X-ray” in medical application. We also presented a summary of Caeleste’s research and development in CMOS circuitry for photon counting color X-ray sensors.

Caeleste photon counting technology makes use of standard CIS technology and works with off-the-shelf scintillators. It is portable to any foundry where pinned diode module is available. Caeleste sensor architecture is designed to maximize yield of wafer-scale devices.

Although CMOS X-ray sensor for large scale medical application will remain more expensive than TFT plates, this photon counting technology provide tangible added value for the diagnostician at the most affordable manufacturing cost.

Contrary to TFT sensors, this CMOS chip is straightforward to interface: a simple USB interfaced camera based on the 92x90 pixels demonstrator is under development. It will real-time evaluation on site on various equipment.

Beyond medical application, this technology is also usable as such in other X-ray applications such as non-destructive testing, luggage inspections, etc.
### Annex 1: Demonstrator specifications

<table>
<thead>
<tr>
<th>Item</th>
<th>Specification</th>
<th>Comment, expected performance</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Geometrical &amp; technological</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Technology</td>
<td>0.18µm</td>
<td>Standard CMOS</td>
</tr>
<tr>
<td>Starting wafer material concept</td>
<td>High resistivity EPI</td>
<td></td>
</tr>
<tr>
<td>Pixel size pitch (µm)</td>
<td>100µm</td>
<td>Scalable down to 50µm</td>
</tr>
<tr>
<td>Transistor count per pixel</td>
<td>45 T</td>
<td>27 T for Black &amp; White version</td>
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<tr>
<td>Photon sharpening</td>
<td>No</td>
<td></td>
</tr>
<tr>
<td>Color X-ray</td>
<td>Yes</td>
<td>Two colors</td>
</tr>
<tr>
<td>Array size</td>
<td>92 x 90 pixels</td>
<td>Of which the right most column is a column with variants and pixels with electrical inputs.</td>
</tr>
<tr>
<td>IC size</td>
<td>1x1cm</td>
<td>Between bondpad extremities</td>
</tr>
<tr>
<td>Reference pixels</td>
<td>-the array contains 3 pixels with partial metal cover for test purposes</td>
<td></td>
</tr>
<tr>
<td></td>
<td>-in the right column only: pixels externally connected with electrical input</td>
<td></td>
</tr>
<tr>
<td>Variant pixels</td>
<td>About 16 variants in the right column</td>
<td></td>
</tr>
<tr>
<td>Windowing (region of interest, random addressing)</td>
<td>Random access in X and Y</td>
<td>Variants play with geometry and implants layers</td>
</tr>
<tr>
<td><strong>Electrical</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Analog photon counting C/C ratio</td>
<td>300:1 or slightly above</td>
<td>This low value was deliberately chosen for easy testability. In later devices this will be increased to &gt;1000:1</td>
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<tr>
<td>Full readout frame rate</td>
<td>30fps</td>
<td>Depends actually form addressing rate.</td>
</tr>
<tr>
<td>Wavelength spectrum (nm)</td>
<td>Typical Si</td>
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<tr>
<td>Spectral response (SR)</td>
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<td></td>
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<tr>
<td>QE*FF</td>
<td>High, not separately measured</td>
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<tr>
<td>packet size threshold [e-]</td>
<td>100e-</td>
<td></td>
</tr>
<tr>
<td>Packet size difference discrimination</td>
<td>15e_RMS</td>
<td></td>
</tr>
<tr>
<td>Q_N [e_RMS]</td>
<td>15e_RMS</td>
<td>Threshold noise</td>
</tr>
<tr>
<td>Maximum count rate (separating two pulses)</td>
<td>At least 300kHz</td>
<td>Under electrical input. Higher values can be obtained by tradeoff between speed/power/noise</td>
</tr>
<tr>
<td>Thresholds of comparators</td>
<td>Adjusted by voltage</td>
<td>two channels per pixel</td>
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<td><strong>Environmental and packaging</strong></td>
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<td>Specification temperature range</td>
<td>20…30°C</td>
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</tr>
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<td>Operation temperature range</td>
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<td>non-operational</td>
<td>-40…+60°C</td>
<td>Depends actually on package and scintillator</td>
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<td>item</td>
<td>specification</td>
<td>Comment, expected performance</td>
</tr>
<tr>
<td>-----------------------------------------</td>
<td>-------------------------------------------------------------------------------</td>
<td>--------------------------------</td>
</tr>
<tr>
<td>temperature range</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Interface, driving</td>
<td>Standard logic level, 0-3.3 V input, medium-speed analog output</td>
<td></td>
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<tr>
<td>Number of supplies</td>
<td>5 domains</td>
<td>Single voltage used. 3.3V-0.0V (up to 3.6V-0.0V). multiple separate analog/digital pairs</td>
</tr>
<tr>
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<td>100 PGA or Chip on board</td>
<td>43 are used for the array itself 51 for standalone test-structures on the same die</td>
</tr>
<tr>
<td>ESD protection</td>
<td>On all IO</td>
<td></td>
</tr>
<tr>
<td>Power consumption</td>
<td>20mW</td>
<td></td>
</tr>
<tr>
<td>Output channels</td>
<td>2</td>
<td>2 comparators or 2 counters read out in parallel</td>
</tr>
</tbody>
</table>
Annex 2: References