A wafer-scale CMOS APS imager for medical X-ray applications

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Abstract

This paper presents a wafer-scale 77.3mm x 145mm 3-side buttable CMOS APS image sensor intended for use as one imager tile of an X-ray mammography detector. The final 232 mm x 290 mm detector comprises 2x3 butted CMOS tiles, a fiber optic plate with scintillator, and readout electronics. The CMOS imager design targets, architecture and evaluation results are presented.

Introduction

Most of the currently installed medical digital X-ray imaging systems use either CCD-based image-intensified cameras with complex optics or compact lens-less flat panel detectors that use amorphous silicon (a-Si) technology in conjunction with conversion media like CsI (indirect conversion) or a-Se (direct conversion). Image intensified solutions suffer from poor optical collection efficiency, low resolution, poor image geometry and inconvenient form factors. Amorphous silicon-based flat panel detectors overcome many of these obstacles, but suffer from high noise floor, image lag and low read-out speed, and are limited to large pixel pitches. Both solutions lead to unacceptable performance compromises in low-dose and high-speed applications. To overcome these limitations we have designed a 3-side buttable, 77.3mm x 145mm, wafer-scale CMOS APS imager with a 33.55µm x 33.55µm pixel pitch. We have built a 3x2 butted array of these sensors to achieve a panel dimension of 232 mm x 290mm that is compatible with the requirements for Full-Field Digital Mammography (FFDM) in screening and biopsy modes.

Image Sensor Tile Development

The image sensor requirements were derived from the X-ray mammography application. Typical mammography application conditions are summarized in Table 1. Assuming the use of a Fiber-Optic Plate (FOP) with a CsI scintillator, this translates to the CMOS image sensor requirement specification as summarized in Table 2.

Mammography requirements	**Specification**
Image size (FFDM)	23 x 29 cm^2
Resolution (DQE)	> 10 % @ 11lp/mm
Pixel pitch (biopsy / spot)	25..50 µm
Pixel pitch (screening)	50..70 µm
Frames per second (biopsy)	> 5..10
Frames per second (screening)	> 1
Dynamic Range	> 70 dB
Non-Linearity	< 2 %
Max. column / row defect	< 2 pixels
Mammography conditions	**Specification**
Average X-ray dose detector surface	70 µGy, 20..35 kVp
X-ray exposure time	0..2 sec
Competitive features	**Comment**
Dose Sensing	Patient dose reduction
Mode switching biopsy /screening	Multi-functionality
Radiation tolerance (End-of Life, at sensor surface)	100 Gy

Table 1. Typical mammography application conditions

The sensor tiles were made on 8” wafers in a 0.3 µm CMOS imaging technology. Each tile has 2304 x 4320 pixels and 4 outputs designed for 40MHz pixel frequency. Since a single tile is larger than the maximum field-of-view of the stepper, stitching is used to combine four identical image sensor slices together to form one wafer-scale sensor tile of 77.3mm x 145mm. Figure 1 shows the schematic for a single slice. Each slice contains all the functionality required for the pixel array control and read-out.

Sensor(tile) performance	Specification
Resolution (tile)	2304 (H) x 4320 (V)
Pixel size	33.55µm x 33.55 µm
Fill factor	84 %
Full Well Capacity	650 ke-
Conversion Gain	2.45 µV/e-
Noise Floor	175 e-
Dynamic Range	71.4 dB
Dark current @ 40 Celsius	44 pA/cm^2
Non-linearity, 10..90% FS	~ 1 %
Radiation tolerance	100 Gy
Image lag	< 0.1 %
Dose Sensing density	1 pixel / 0.54 mm^2
Frame Rate - Full resolution - 2x2 binning - dose sensing	 8.7 FPS 14.8 FPS 5000 FPS

Table 2. Sensor requirement specification

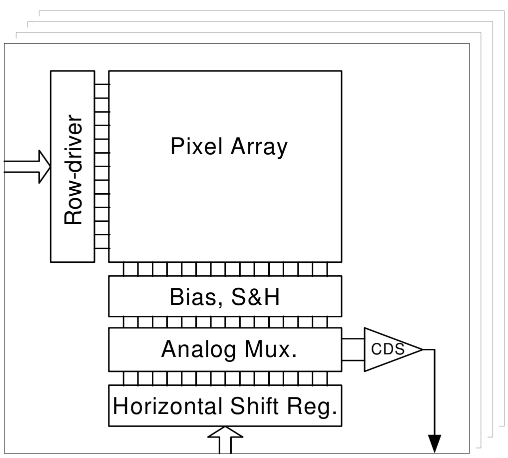

Figure 1. Single slice sensor architecture

The sensor has an option for high-speed non-destructive image read-out of a predefined set of 'dose-sensing' pixels. This feature supports fast Auto Exposure Control (AEC) algorithms which can be used to achieve best diagnostic results at lowest X-ray dose. During the full resolution image data read-out, the dose-sensing pixels are treated as 'normal' pixels. Figure 2 depicts the organization of the dose-sensing pixels over the pixel array and shows that each column of a selected array unity block contains a single dose-sensing pixel. Via a separate hard-wired addressing method all dose-sensing pixels in the unity-block can be addressed at once, which supports the read-out of these pixels in a single line-time period. The device pixel array is build from an array of 4 x 9 unity blocks stitched together during the chip production process. Consequently all device dose-sensing pixels can be read non-destructively in only 9 line-time periods. At 40 MHz pixel speed this results in a 187 µsec read-out period for a single dose-sensing frame.

Since minimum X-ray exposure periods are in the range of a few milliseconds, an accurate AEC algorithm can be developed to stop the X-ray source when a high quality image is integrated in the pixel array, resulting in a minimized X-ray dose.

The density of the dose-sensing pixels is 1 pixel per 0.54 mm2, sufficient for accurate exposure control.

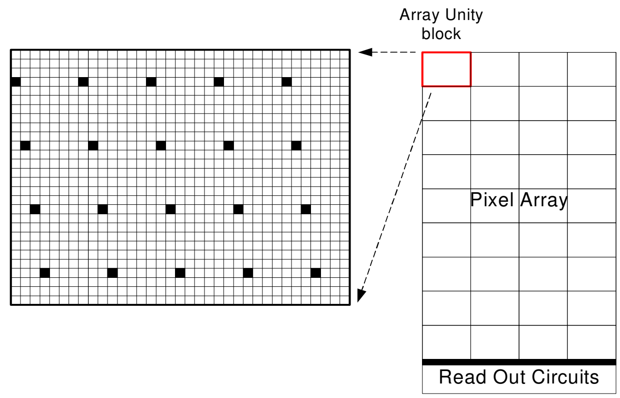

Figure 2. Organisation of dose-sense pixels

The row-driver is merged into the pixel array to minimize the light insensitive gap in between sensors in case of a butted assembly, figure 3. The embedded row-driver is distributed over multiple adjacent columns to avoid dead columns by design since the width of a row-driver exceeds the width of a single pixel. This reduces the fill factor of the affected pixels from 82% to values between 62% and 74%.

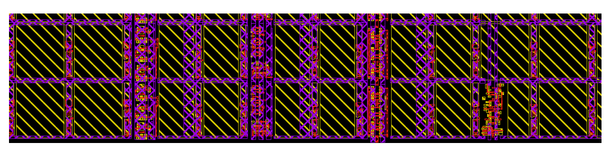

Figure 3. Distributed row-driver design

The reduced sensitivity is in first order compensated by increasing the conversion gain of the pixels having row-driver functionality. The remaining sensitivity differences can be easily corrected in the application by gain calibration, since the spectral distribution of the light is always identical (defined by the scintillator).

Figure 4 depicts the design of the embedded three-transistor pixel. The pixel uses a pinned photodiode to ensure low dark current, even over extended dose range (radiation-hard design). The high full well capacity is achieved through use of an in-pixel MOS capacitor. The pixel has a hardware programmability option for the pixel select-signal with a single metal-metal VIA. For dose-sensing pixels the "DS_select" signal is connected with the select gate to support simultaneous selection of the dose-sensing pixels per array unity block.

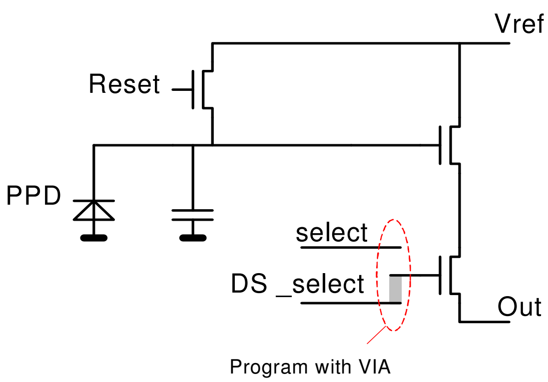

Figure 4. Pixel architecture

The distance between the edge of the outer array pixels to the edge of the silicon (after dicing) is approximately 20µm.

Evaluation Results

Figure 5 shows an assembled tile without FOP + scintillator in a protective box used for evaluating the sensor design.

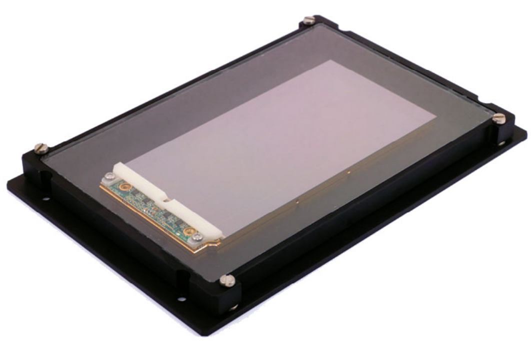

Figure 5. Assembled sensor tile in protective box

The measured saturation charge is 650 ke-. The conversion factor is 2.45 µV/e-. Figure 6 shows the device overall non- linearity to be better than 1.5 % over the range from 10 to 90 percent saturation swing. Below 10 % saturation swing the device shows a nice linear response as depicted in figure 7. Dark current measurements showed an average dark current of 44 pA/cm2 at 40 Celsius before exposure to ionization dose. Figure 8 depicts the dark current increase as a function of Total Ionization Dose to the bare silicon.

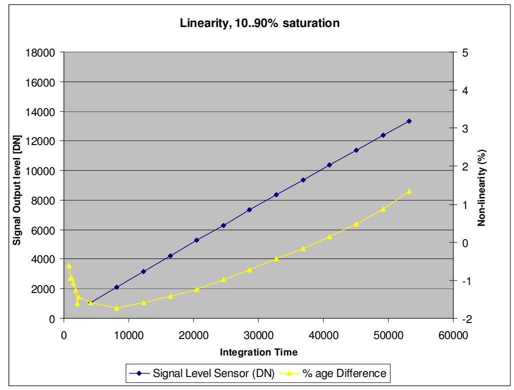

Figure 6. Non-linearity, 10..90% saturation swing

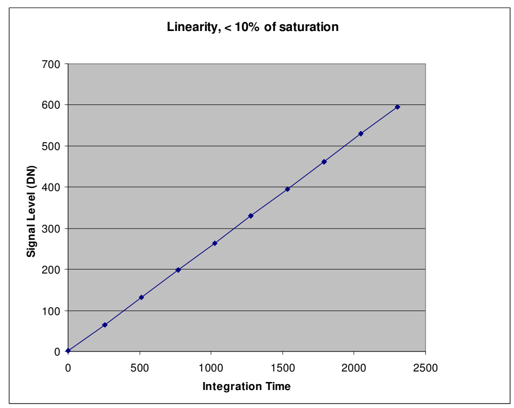

Figure 7. Non-linearity, < 10% saturation swing

The dose was applied to the silicon under continuous X-ray exposure. Repeated measurements have shown that annealing effects play a role which reduces the dark current over time in case no ionization exposure is applied.

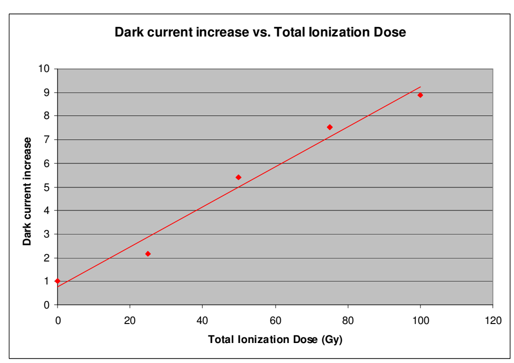

Figure 8. Dark current versus Total Ionization Dose

Figure 9 depicts the achieved Quantum Efficiency over the range of 400 to 700 nm wavelength (including fill-factor effect).

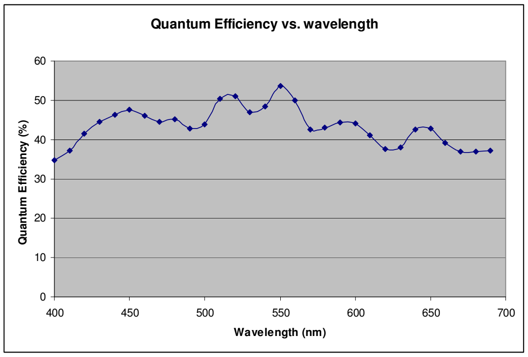

Figure 9. Quantum Efficiency

A Data Double Sampling (DDS) read-out method is used to read the pixel data from the pixel array. The kTC noise introduced by this read-out method is in the range of 400μV. The overall noise-floor was measured to be 430μV, which equals ~175 electrons. This means that the noise contribution of the analog chain is in the range of 170μV.
The initial X-ray evaluation was done on single tiles with attached FOP with CsI scintillator. The measured Detective Quantum Efficiency (DQE) was compared against the DQE of a state-of-the-art a-Si panel, Figure 10 [2]. The results clearly show the superior performance at low dose for the CMOS APS based detector.

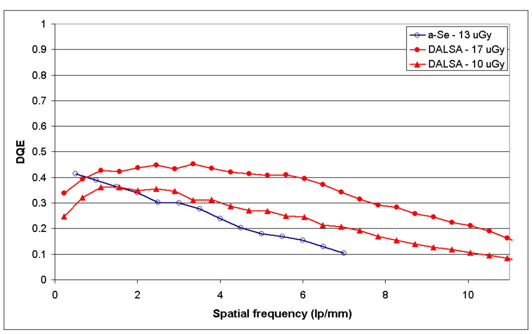

Figure 10 Detective Quantum Efficiency

At 40MHz pixel rate per output, a full resolution frame-rate of 8.7 fps is obtained; with 2 x 2 binning, this increases to 14.8 fps. Finally figure 11 shows the 'first patient result' obtained on a total detector made of 2x3 butted tiles with FOP + scintillator as depicted in figure 12.

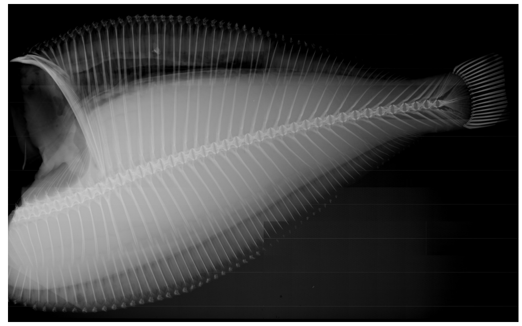

Figure 11. "First patient" results

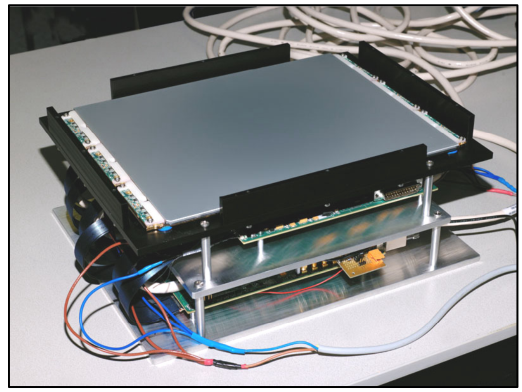

Figure 12. 2x3 butted sensor assembly with FOP + scintillator

Conclusions

A 3-side buttable radiation-hard 8" wafer-scale CMOS image sensor for use in an X-ray mammography detector was developed. The comparatively low noise of the CMOS imager contributes to a higher DQE at lower X-ray dose. The negligible lag and high frame-rates make these tiles suitable for use in large panels intended for dynamic medical X-ray applications. Distributed row-drivers and dedicated edge pixel design allow a butting gap of less than 70μm. Dose-sensing can easily be included to reduce the patient radiation dose.

Acknowledgements

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References

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[2] L.K. Chueng et al, "Image performance of a new amorphous selenium flat panel x-ray detector designed for digital breast tomosynthesis",
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