High Dynamic Range CMOS-based Mammography Detector for FFDM and DBT

Inge M. Peters[1*], Chiel Smit[1], James J. Miller[2], Andrey Lomako[2]
[1]Teledyne DALSA Professional Imaging, High Tech Campus 27, 5656 AE Eindhoven, The Netherlands, [2]Teledyne DALSA Professional Imaging, 605 McMurray Rd., Waterloo, Ontario, Canada

ABSTRACT

Digital Breast Tomosynthesis (DBT) requires excellent image quality in a dynamic mode at very low dose levels while Full Field Digital Mammography (FFDM) is a static imaging modality that requires high saturation dose levels. These opposing requirements can only be met by a dynamic detector with a high dynamic range. This paper will discuss a wafer-scale CMOS-based mammography detector with 49.5 µm pixels and a CsI scintillator. Excellent image quality is obtained for FFDM as well as DBT applications, comparing favorably with a-Se detectors that dominate the X-ray mammography market today.

The typical dynamic range of a mammography detector is not high enough to accommodate both the low noise and the high saturation dose requirements for DBT and FFDM applications, respectively. An approach based on gain switching does not provide the signal-to-noise benefits in the low-dose DBT conditions. The solution to this is to add frame summing functionality to the detector. In one X-ray pulse several image frames will be acquired and summed. The requirements to implement this into a detector are low noise levels, high frame rates and low lag performance, all of which are unique characteristics of CMOS detectors. Results are presented to prove that excellent image quality is achieved, using a single detector for both DBT as well as FFDM dose conditions.

This method of frame summing gave the opportunity to optimize the detector noise and saturation level for DBT applications, to achieve high DQE level at low dose, without compromising the FFDM performance.

Keywords: CMOS X-ray detectors, mammography, Digital Breast Tomosynthesis, Full Field Digital Mammography, low dose, frame summing

1. INTRODUCTION

Recently several studies have been published showing the benefits of CMOS-based X-ray detectors for Digital Breast Tomosynthesis (DBT)[1,2]. The advantage of CMOS active pixel sensors compared to thin-film transistor (TFT) passive pixel sensor technology are the low electronic noise, the faster frame rate and the smaller pixel pitch[3]. The electronic noise of the CMOS sensor presented in this paper is below 200 e^- [4], whereas TFT detectors for mammography typically demonstrate significantly higher electronic noise, which could reach 1000 e^- and even higher. The reduction in pixel size for TFT technology will come at the cost of fill factor, while the 49.5 µm CMOS sensor presented in this paper has a fill factor of 79%. Small pixel pitch and low noise levels are needed for the detection of small microcalcifications[5,6]. To diagnose breast cancer in an early stage, the detection of microcalcifications smaller than 200 µm is crucial. Park et al.[7] showed that using an X-ray detector with a ~75 µm pixel 165 µm calcifications could be resolved. With a pixel of around 50 µm, the detection of microcalcification of ~100 µm becomes feasible.

Full Field Digital Mammography (FFDM) requires higher saturation dose levels than DBT mode. To combine both modes into one detector, a high dynamic range is required. This can be achieved by designing two full well capacities

[*]inge.peters@teledynedalsa.com

into the CMOS sensor to achieve two saturation dose levels[8,9], one full well capacity optimized for DBT and the other for FFDM applications. In this paper, a different approach called frame-summing is presented. In one X-ray pulse several image frames are acquired and summed. The advantage of this method is that the saturation dose level for FFDM mode is adjustable through the number of frames summed.

2. METHODOLOGY

The 23x29 cm^2 mammography detector consists of four wafer-scale CMOS sensors having a small 49.5x49.5 µm^2 pixel, and an on-chip 14-bit analog-to-digital converter with low readout noise and low power consumption. The detector operates at 8 frames per second in full resolution and 16 fps in 2x2 pixel binning mode. A CsI scintillator tailored for mammography applications converts the X-rays to visible light.

Typically, the saturation dose requirement of a mammography detector is set by the FFDM application. If the detector should also serve the DBT application, a different dose range (typically 10% of the FFDM dose) needs to be mapped onto the full ADC range. To achieve this, a gain switch is often applied. Consequently, only a fraction of the available signal storage capacity of the pixels is used and the signal-to-noise performance decreases. This way of combining FFDM and DBT in one detector is schematically illustrated in Figure 1. Considering the low dose levels in DBT, the reduction in signal-to-noise ratio is a severe drawback of gain switching.

We present a method that overcomes this drawback and that delivers superior signal-to-noise performance for both FFDM and DBT in one detector. The detector described in this paper has a pixel saturation dose of about 1 mGy and is optimized for the more demanding DBT application. In FFDM mode, the detector is read out several times during the X-ray exposure and the frames are summed to obtain the final FFDM image. The pixels will not saturate due to the high frame rate achievable with the CMOS technology. The signal-to-noise performance improves because the summing operation averages the noise and increases the output bit depth. This frame summing technology is schematically illustrated in Figure 2.

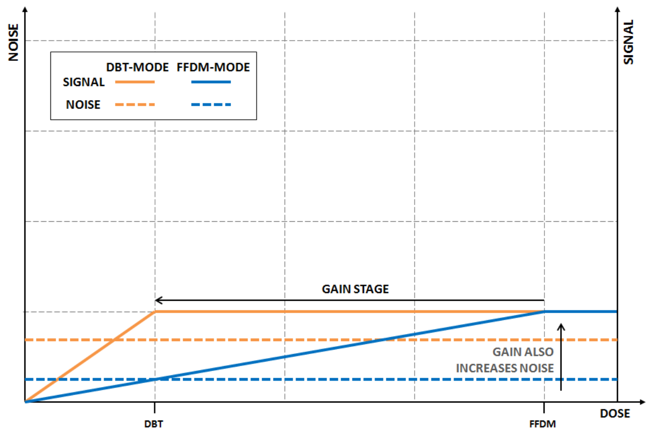

Figure 1 FFDM & DBT modes using gain switch

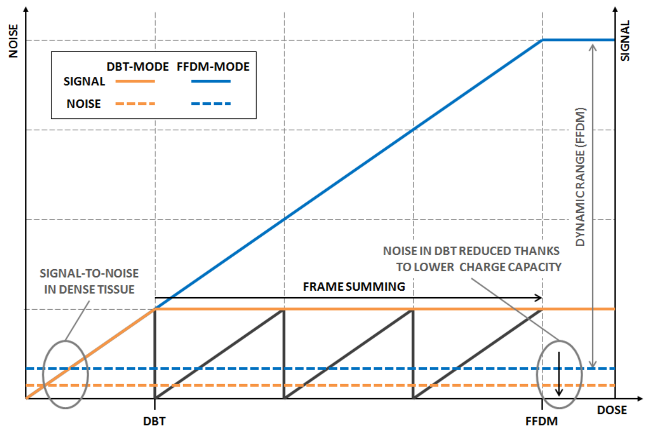

Figure 2 FFDM & DBT modes using frame summing

To maintain excellent X-ray image quality by this implementation, several requirements need to be fulfilled. The frame rate in combination with the saturation level must accommodate the maximum application dose rates. Our detector is designed for a maximum 10 mGy/s dose rate in full resolution mode, which is sufficient for FFDM applications. The detector response versus dose should be very linear to avoid artifacts. It is not allowed to lose any X-ray dose during frame-summing operation. This is achieved by operating the CMOS detector in continuous rolling shutter mode such that the photodiode is always integrating. The noise level of the detector needs to be low and the frame rate setting has to be optimized for the dose rate, such that each single frame avoids saturation in any part of the image.

3. RESULTS

Linearity

In Figure 3 the linearity of the mammography detector in DBT mode, and in FFDM mode is shown. The measurements were done at RQA-M2 radiation quality (Mo/Mo, 28 kV, 2 mm Al). In DBT mode, the X-ray pulse was exposed within a single frame capture. The integration time of the detector was 923 ms and the maximum output level was 16384 DN, which explains the deviation of the last measurement point in the graph.

In FFDM mode, four frames are acquired in sequence and then summed within the detector, and the single summed image is the output. The integration time per frame was 250 ms resulting in a frame rate of 4 fps. The maximum output level of ~56000 DN, is somewhat less than 4x16384 DN, because of the continuous rolling shutter readout mode. In this readout mode, the detector is read out in a row by row sequence while all rows except the one read out are continuously integrating. The readout time of a single row is very short 42 µs and does not result in any meaningful loss in X-ray dose. The readout of one complete frame, takes about 123 ms. To capture an X-ray pulse requires that the pulse has to end before the start of the last frame readout, thereby reducing the integration time of the last frame by 123 ms. The saturation dose in FFDM mode of this detector depends on the number of frames summed and could be adjusted to specific application requirements. A sequence of any number of frames can be captured and summed.

Both, DBT mode and FFDM mode show excellent linearity. Summing frames has no impact on the achieved linearity graph. The sensitivity in DBT and FFDM modes is also identical showing that there is no loss in X-ray dose. At low exposure conditions, small inaccuracies in dose measurements will result in larger deviations from linearity in percentage terms.

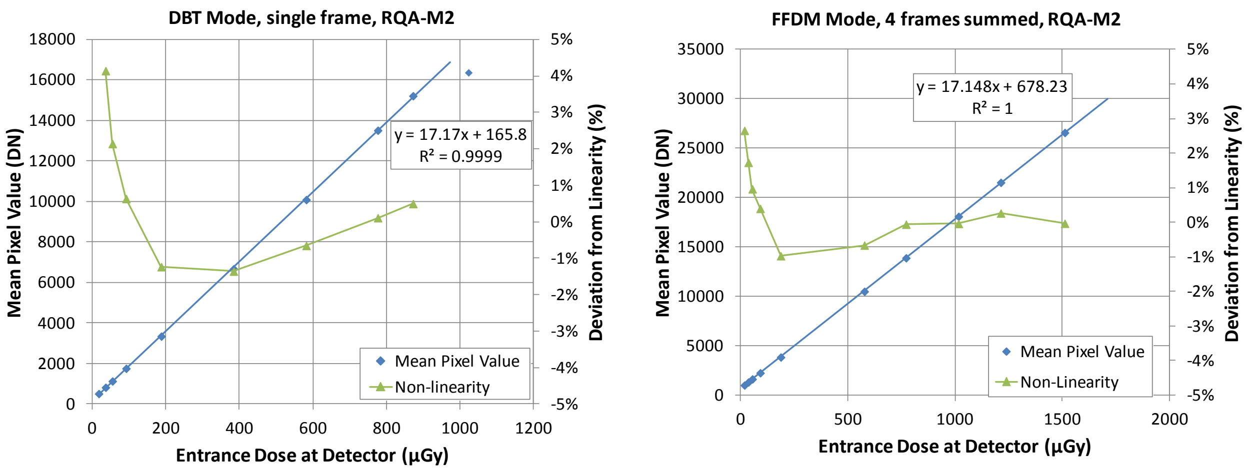

Figure 3 Blue curve shows the sensitivity in DBT (left) and FFDM (right) mode measured at RQA-M2. Green curve shows the deviation from a linear fit through the measurement points.

MTF

In Figure 4 the MTF evaluation of the detector is shown. Measurements were done according to the IEC 62220-1-2 standard at RQA-M2, 280 µGy. The MTF performance is lower than state-of-the art a-Se direct conversion detectors (at 5lp/mm MTF=35% vs MTF=50% for a-Se[10]), but better than other CMOS-based mammography detectors (MTF at 5lp/mm = 30%)[1,2].

DQE

In Figure 5 the DQE performance in DBT mode and in FFDM mode is shown. The DQE was measured according to the IEC 62220-1-2 standard at RQA-M2. For DBT and FFDM mode, the same detector settings as for the linearity measurements were used. The DQE in DBT mode was measured at two dose levels, a normal dose of 285 µGy and a low dose of 27 µGy at the detector entrance surface. For FFDM mode the DQE was measured at 285 µGy and 2 mGy at the detector entrance surface.

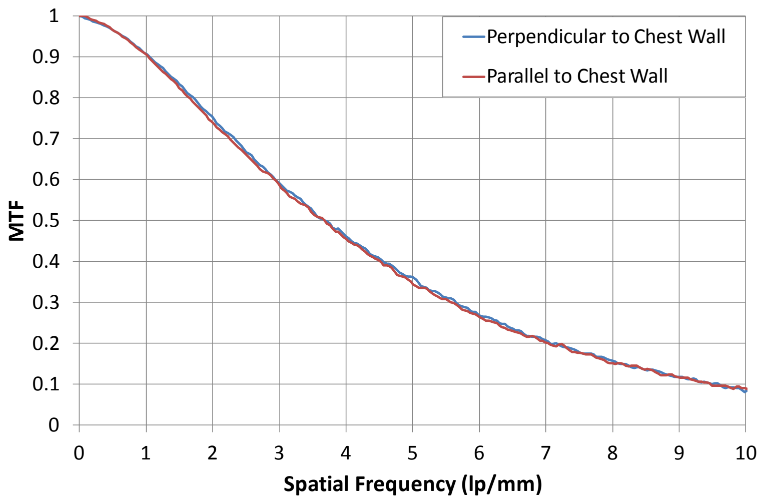

Figure 4 The MTF performance at RQA-M2, 285 uGy.

The DQE performance for DBT and FFDM mode at the same X-ray exposure of 285 µGy is similar. For DBT mode, this DQE was obtained with a single image capture while for FFDM, 4 frames were summed in the detector. At high dose levels, above saturation level in DBT mode, the DQE performance in FFDM mode is the same at 285 µGy and shows that summing does not impact the performance. This graph shows that the detector delivers good image quality in both modes and that it outperforms competitive detectors especially at higher spatial frequencies[1,2,10]. Furthermore, the excellent DQE performance in both DBT and FFDM modes has been confirmed by imaging experiments on the CDMAM phantom.

Measurements of DQE using RQA-M2 beam quality at very low dose conditions were not available and to show the low noise performance of the CMOS detectors a model[11] was used. It has been shown that this modeling predicts the low dose performance of CMOS-based X-ray detectors very well for different optical stack configurations and different noise and saturation dose levels[9]. In Figure 6 the calculated low dose performance (green curves) at 5 µGy and 10 µGy are plotted based on the measurements at 285 µGy. For comparison, the measured and calculated performance at 27 µGy is shown. The DQE performance starts to decrease below 20 µGy, but at 5 µGy the DQE is still larger than 35% at 5lp/mm.

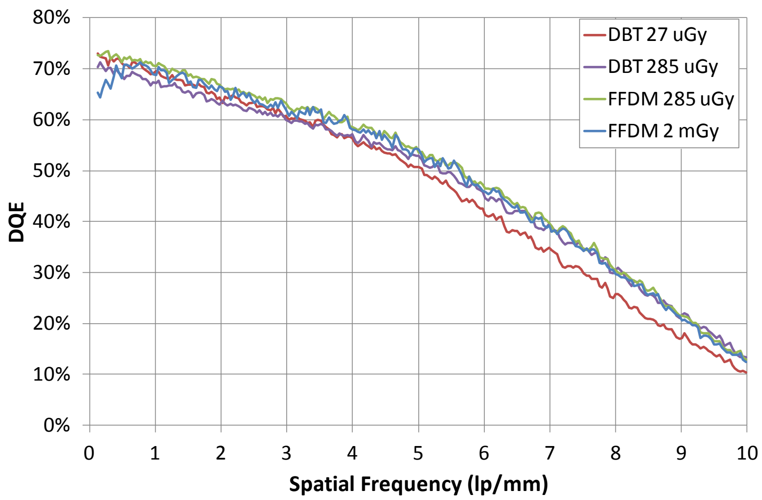

Figure 5 DQE performance at RQA-M2 in DBT mode (single image capture) and FFDM mode (4 frames summed) at typical dose conditions.

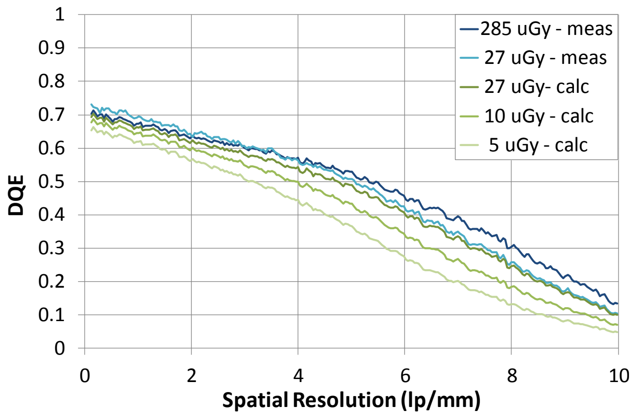

Figure 6 The measured (blue) and modeled (green) DQE performance in DBT mode at RQA-M2.

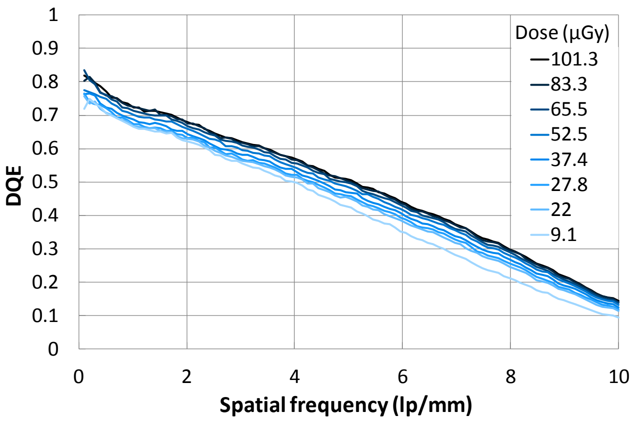

Figure 7 The measured DQE performance as function of dose in DBT mode at 28 kV, W/Rh.

In Figure 7 the DQE as function of dose is plotted using a different mammography spectrum. For this measurement a W/Rh anode/filter combination was used at 28 kV with an added 2 mm Al filter. For this spectrum higher DQE values are achieved and the DQE does not start to decrease until 10 µGy.

Lag and Ghosting

For good image quality in DBT mode, lag and ghosting are critical parameters. Lag is the residual signal in a dark image generated by previous X-ray exposures. The lag was measured with the detector operated in rolling shutter mode at 10.5fps. The measurements were executed using a W-anode, 40 kV, and 2 mm Al filter. After an X-ray pulse of 3 mGy with a duration of 2s the residual signal was measured in dark, i.e. without any X-ray exposure. The result is plotted in Figure 8. After 0.1s the lag is already smaller than 0.17%. This is significantly better than reported for a-Se detectors[12] and demonstrates a valuable advantage of CMOS detector technology.

Ghosting is the change in X-ray sensitivity as a result of prior radiation exposure. The ghosting was measured by first shielding half of the detector with lead and applying a 15 mGy exposure to the detector. The ghosting is defined as the sensitivity difference between the irradiated regions relative to the regions covered with lead. In Figure 9 the ghosting result is plotted. The measurement shows that the ghosting is 0.16% after 90s and 0.08% after 5 minutes. The lag and ghosting performances are dominated by the CsI scintillator.

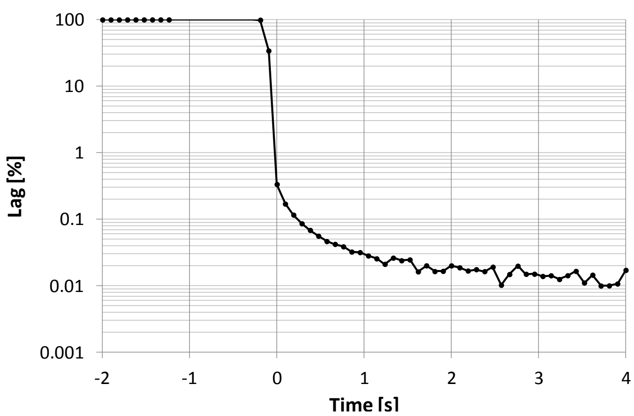

Figure 8 Lag after 3 mGy X-ray exposure during 2s.

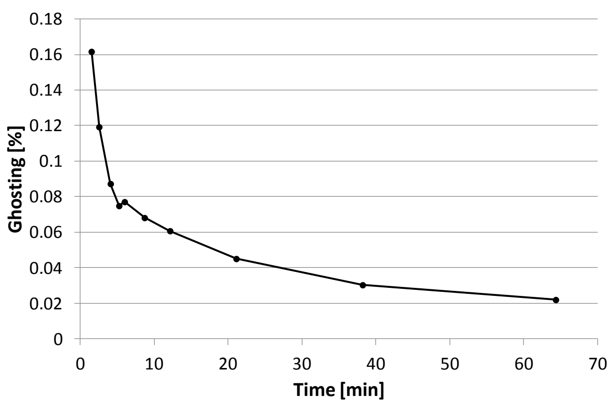

Figure 9 Ghosting after covering half of the detector with lead and irradiating with15 mGy X-ray exposure.

4. CONCLUSIONS

The characteristics and performance of a high dynamic range CMOS X-ray mammography detector developed to support both DBT and FFDM applications have been presented, showing that it offers excellent image quality for both applications and compares favorably with state-of-the art a-Se mammography detectors. At higher spatial frequencies it outperforms in DQE performance providing the opportunity to detect microcalcifications of ~100 µm. The high dynamic range is achieved by the implementation of frame summing. This feature also offers the possibility to implement advanced patient dose management functionality during the procedure. For example, the X-ray pulse duration needed to achieve a certain dose in a specific region of the image can be calculated from the first frame of the image, so that the dose applied to the patient will be optimized for the clinical imaging task.

5. ACKNOWLEDGEMENT

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REFERENCES

[1] Zhao, C., Kanicki, J., Konstantinidis, A.C. and Patel, T., "Large area CMOS active pixel sensor x-ray imager for digital breast tomosynthesis: Analysis, modeling, and characterization," Med. Phys. 42 (11), 6294-6308 (2015)

[2] Zhao, C., Konstantinidis, A.C., Zheng, Y., Anaxagoras, T., "50 µm pixel pitch wafer-scale CMOS active pixel sensor x-ray detector for digital breast tomosynthesis," Phys. Med. Biol. 60 (2015), 8977–9001 (2015)

[3] Zentai, G., "Comparison of CMOS and a-Si flat panel imagers for X-ray imaging," Proc. IEEE IST 05/211, 194-200 (2011)

[4] Weisfield, R.L., Bennett, N.R., "Electronic noise analysis of a 127 µm pixel TFT/photodiode array," Proc. SPIE 4320, 209-218 (2001)

[5] Burgers, A.E., "Effect of Detector Element Size on Signal Detectability in Digital Mammography," Proc. SPIE 5745, 232-242 (2005)

[6] Wheeler, F.W., Perera A.G.A., Claus, B.E., Muller S.L., Peters G., Kaufhold, J.P., "Micro-calcification detection in digital tomosynthesis mammography," Proc. SPIE 6144, (2006)

[7] Park, H.S., Kim, Y.S., Kim, H.J., Choi Y.W., Choi J.G., "Optimization of configuration parameters in a newly developed digital breast tomosythesis system," J. of Radiat. Res 55, 589-599 (2013)

[8] Konstantinidis, A.C., Szafraniec, M.B., Speller, R.D., Olivo A., "The Dexela 2923 CMOS X-ray detector: A flat panel detector based on CMOS active pixel sensors for medical imaging applications", Nuclear Instruments and Methods in Physics Research A 689, 12–21 (2012)

[9] Maes, W.H., Peters I.M., Smit, C., Kessener, Y., Bosiers, J., "Low-dose performance of wafer scale CMOS-based X-ray detector," Proc. SPIE 9412, (2015)

[10] http://www.analogic.com/products-medical-digital-mammography.htm

[11] Granfors, P.R., Aufrichtig R., Possin, G.E., Giambatista B.W., Huang Z.s., Jianqiang, L., Bing, M., "Performance of a 41x41 cm2 amorphous silicon flat panel x-ray detector designed for angiographic and R&F imaging applications," Med. Phys. 30 (10), 2715-2725 (2003)

[12] Zhao B., Zhao W., "Imaging performance of an amorphous selenium digital mammography detector in a breast tomosynthesis system," Med. Phys. 35 (5), 1978-1987 (2008)