Hybrid Direct Conversion Detectors for Digital Mammography[1]

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Abstract

Hybrid pixel detector arrays that convert x-rays directly into charge signals are under development at NOVA for application to digital mammography. This technology also has wide application possibilities in other fields of radiology and in industrial imaging for applications in nondestructive evaluation and inspection. These detectors have potentially superior properties compared to either emulsion based film, which has non-linear response to x-rays, or phosphor-based detectors in which there is an intermediate step of x-ray to light photon conversion[1]. Potential advantages of direct conversion detectors are high quantum efficiencies (QE) of 98% or higher (for 0.3 mm thick CdZnTe detector with 20 keV x-rays), improved contrast, high sensitivity and low intrinsic noise. These factors are expected to contribute to high detective quantum efficiency (DQE). The prototype hybrid pixel detector developed has 50 x 50 microns pixel size, and is designed to have linear response to x-rays, and can support a dynamic range of 14 bits. Modulation Transfer Function (MTF) is measured on a 1-mm silicon detector system where 10% or better modulations are obtained at 10 lp/mm spatial frequency. Preliminary DQE measurements of the same system yields a value of 55% at zero spatial frequency.

In this paper, we report data of our first full size prototype readout ASIC chips hybridized with both silicon and CdZnTe detector arrays and present initial MTF and DQE measurement results as well as some test images.

I. INTRODUCTION

Digital mammography offers the potential of improved image quality and the possibility of increased performance of detecting breast cancer, particularly in women with dense breasts where current screen-film mammography is often lacking. Digital mammography will also facilitate the implementation of CAD (Computer Aided Diagnosis) and telemammography [1].

The key element in a digital mammography system is the detector, which converts the x-ray pattern transmitted from the breast into electronic signals that can be digitized. Important detector properties are dynamic range, quantum efficiency, sensitivity, contrast, noise, linearity and the ability to provide high spatial resolution over the required detection area.

Current detector approaches for digital mammography usually employ phosphors which, in response to x-ray absorption, produce light photons which are then converted into an electronic signal. This process is inefficient and can lead to increased image noise, particularly when signals are low.

Our approach is to avoid the problems associated with the phosphors by using direct conversion solid state detectors. In these detectors x-ray photons directly create electron-hole pairs without producing the intermediate light photons. Such direct conversion detectors require custom-made readout electronics. Since the pixel sizes are small (50 x 50 micron), it is a major challenge to obtain coupling between all of the detector elements and the readout electronics. Indium bump bonding technique has been used to connect the detector pixels to that of the readout chip. In this technology, small indium bumps are first deposited onto each pixel of the detector and that of the readout chip separately, and then the detector and the readout chip are pressed together where the indium bumps are cold-welded together. Special effort has been made to reduce the inter pixel leakage and improve electrode metal contacts, especially for the CdZnTe detectors.[2]

An ASIC readout chip (MARY: MAmmogRaphY) has been developed to read out the charge signals from two-dimensional detectors such as silicon or CdZnTe pixel arrays, which are hybridized to the readout chip.

The silicon pixel detectors have low Z, low noise, and are excellent for low energy x-ray imaging from 5 to 20 keV. The high-Z CdZnTe pixel detectors can increase the x-ray energy range from 20 to 150 keV.

A major advantage of this technology, compared to some active matrix devices such as amorphous silicon, is that the pixel size can be made almost arbitrarily small without losing the "fill factor." Currently the fill factor for devices like amorphous silicon falls off rapidly with decreasing pixel sizes.

Preliminary data characterizing the silicon detector system, as well as first images taken from silicon and CdZnTe hybrid pixel detectors, demonstrate their potential for many medical and industrial applications.

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II. System Design

A. READOUT ASIC CHIP DESIGN

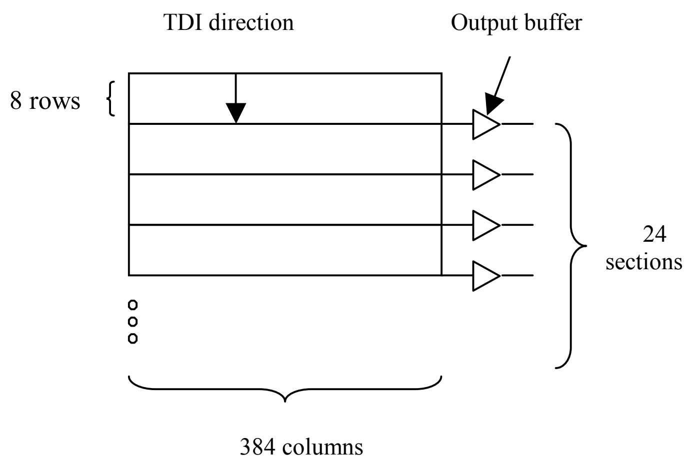

Figure 1. Block diagram of a Large MARY chip.

Figure 1 is a simplified block diagram of the MARY readout chip. Time Delay Integration (TDI) technique has been used in the design.[1] Due to the large dynamic range concern for mammography applications, the 192 total TDI stages are divided into 24 sections and are read out separately. For each section there are only 8 TDI stages before the charge is read out. Each charge well in a pixel has a capacity of about 15 million electrons, and the overall charge capacity reaches 360 million electrons. In addition, a few bad sections can be averaged out with the other working sections. The measured output noise per section is about 7000 electrons, and the overall dynamic range is > 13 bits for the present system. Since the estimated readout chip noise is about 3000 electrons per section, the major contribution of the noise is still from detector dark currents and electronic readout circuitry. Further reduction of the noise by cooling the detectors or redesigning the readout electronics can increase the dynamic range to 14 bits.

The readout chip is designed to handle slot-scan x-rays where the charge transfer speed is about 1 ms per pixel. The charge transfer efficiency (CTE) is measured to be 0.998 per transfer step for the present MARY chip, due to a surface p channel technology selected at the foundry. The CTE could be dramatically increased by selecting a buried channel technology, but at some reduced well capacity.

A high speed data acquisition system has been built to transfer data from the detector into a computer through a serial interconnect cable. The data transfer rate is about 25 Mbytes per second.

B. Detector Design

Two different materials are being used to build the detector hybrids: silicon and CdZnTe.

Silicon has the advantage of easy processing, and the technology for indium bonding silicon chips is mature. Since silicon has an atomic number of 14, it has a low x-ray attenuation coefficient and is expected to have an efficiency of only 63%, for a 1-mm silicon detector for 20 keV x-rays. However, because of the high gain and sensitivity, the DQE will be almost equal to this value. It takes, on average, 3.6 eV to produce one electron-hole pair so that each 20 keV incoming x-ray photon produces about 5,500 electron-hole pairs. The excellent sensitivity and the low statistical variation in the signal per x-ray contribute to high signal to noise ratio (SNR). Both 1 mm and 1.5 mm thick silicon detectors have been tested on a proof-of-principle system.

CdZnTe has several properties that make it potentially useful for digital mammography and other industrial applications. CdZnTe has a high density (5.8 g/cm^3) and a high effective atomic number ($Z_{mean} \approx 50$) which provides good quantum efficiency (>98% at 20 keV for 0.3 mm thick detectors). It also provides good SNR, since each 20 keV x-ray photon can produce about 4,000 electron-hole pairs [3]. There are a few drawbacks associated with this new detector material. The present high-pressure Bridgman technique used to grow this material yields only poly-crystals with defects such as grain boundaries and clusters of high impurity concentrations.[4] It is known that these defects would cause band bending near the surface which can trap the x-ray generated charges and therefore polarize the detector under low temperature or high flux conditions.[4][5] Some deep traps produced by these defects can also reduce the charge transport efficiency. In addition, the hole's mobility-lifetime product is about two orders of magnitude smaller than that of the electrons and, therefore, it is preferable to collect the electrons rather than the holes [6][7].

We also performed preliminary work on a prototype system, using a 2-mm thick CdZnTe detector hybridized to the existing hole-collecting readout chip, in spite of these disadvantages. It is anticipated that the charge collection efficiency will be affected, but valuable information can be obtained which lays ground for our future work where electrons will be collected. A redesign of an electron-collecting version of the MARY chip is in preparation.

Figure 2 shows a photograph of two MARY chip hybrids. The large MARY chip has 384 columns and the small one has 128 columns. Both chips have 192 pixels in the TDI direction.

III. Test Results

With the exception of Figure 8, all results were obtained with the silicon detector system. A 1-mm silicon detector system was used to image a bar pattern under a mammography x-ray machine with the pattern in both TDI scan and non-TDI scan directions. Holes were collected here. Figure 3 shows images where the 9 lp/mm patterns are clearly visible (the Nyquist limit for 50 micron spatial resolution is 10 lp/mm).

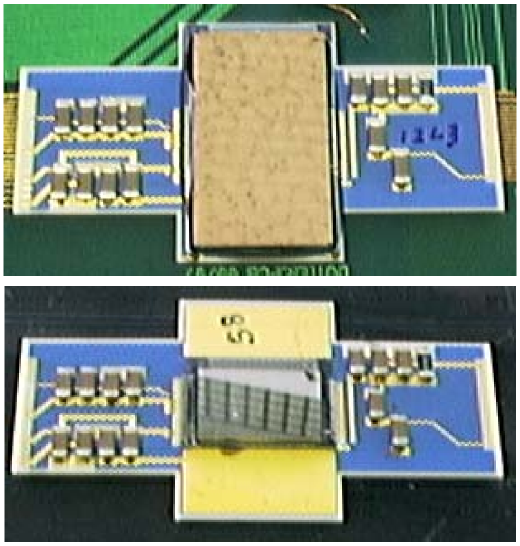

Figure 2: A large MARY chip (Top), hybridized with a 2 mm thick CdZnTe detector, and a small MARY chip (Bottom), hybridized with a 1 mm thick silicon detector, mounted on ceramic chip carriers.

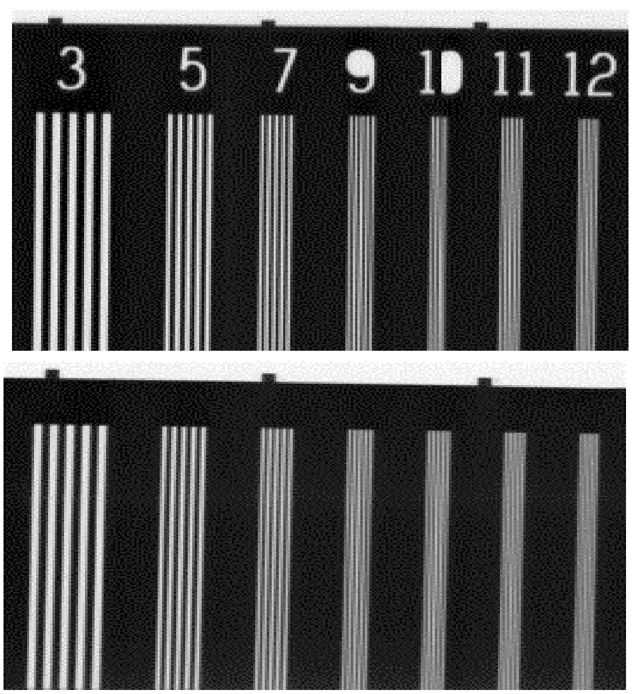

Figure 3: Bar pattern x-ray images using a 1-mm thick silicon pixel detector hybrid. The 9 lp/mm pattern is resolved clearly in the top picture (scan is in the non-TDI direction). In the bottom picture the scan is in the TDI (horizontal) direction.

Figure 4 gives the pre-sampled MTF of the detector in both scan and slot directions. The MTF was determined using a slanted edge technique[8], where a tantalum edge was placed on a large 6.2 mm thick plate of PMMA. The tungsten anode x-ray tube was operated at 30 kVp and 38 mA, with 0.4 mm Al filtration.

The Noise Power Spectra (NPS) was measured using a simulated slit method[9] from 40 images of the uniform PMMA block. The corresponding exposure for each NPS was measured using an ion chamber. Preliminary results indicate that for the best 1-mm silicon detector the DQE reaches 0.55 at zero spatial frequency with 120 mR x-ray exposure.[10]

A 1.5-mm silicon detector hybrid was used to image a jaw phantom. The x-ray machine was set at 90 kV with a tube current of 3 mA. The focal spot was 0.4 mm in diameter and about 50 cm away from the detector. Figure 5 shows the image of a section of the jaw phantom.

Figure 6 is the image of an IC chip taken using the 1.5 mm thick silicon detector hybrid, using an x-ray generator set at 100 kV and 3 mA. The detector bias voltage was 120 V. The 0.025-mm bonding wires of the IC to the carrier are clearly visible in the image.

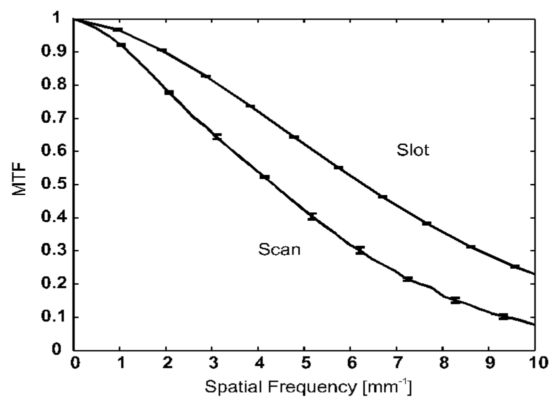

Figure 4: MTF plots of a 1-mm thick silicon pixel detector. The data in both TDI (scan) and non-TDI (slot) directions are shown with standard error with 5 repeated measurements.

Figure 5: Image of a section of a jaw with teeth taken using a 1.5 mm thick hybrid silicon pixel detector.

Figure 7 is an image of a collimator pattern taken using a 1.5-mm thick silicon pixel detector with an x-ray generator set at 49 kV and 40 mA. The focal spot size of the x-ray tube is 0.4 mm in diameter and about 40 cm away from the detector.

A 2-mm CdZnTe pixel detector hybrid (the first prototype which has just been fabricated) was also tested under an x-ray generator set to 160 kV and 2 mA. The focal spot of the tube is 0.2 mm in diameter and about 40 cm away from the detector. The detector bias was set at 600 V. Figure 8 is a picture of the same collimator pattern shown in Figure 7; the smallest holes are 0.15 mm in diameter, shown at the top. The thickness of the metal collimator pattern is 2 mm. Comparing figures 7 and 8, it can be seen that the silicon detector has better spatial uniformity.

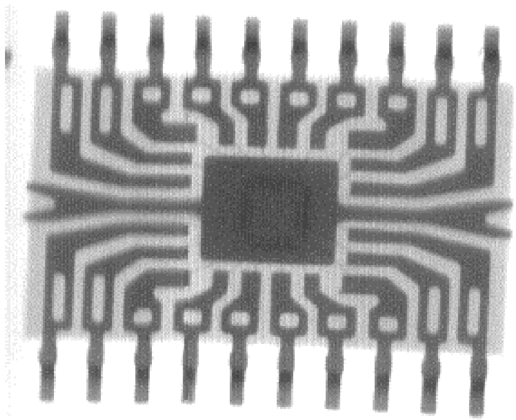

Figure 6: Image of an IC placed at a slight angle to the x-ray scanning direction using a 1.5 mm thick silicon pixel detector. The 0.025-mm diameter wire bonds are clearly visible.

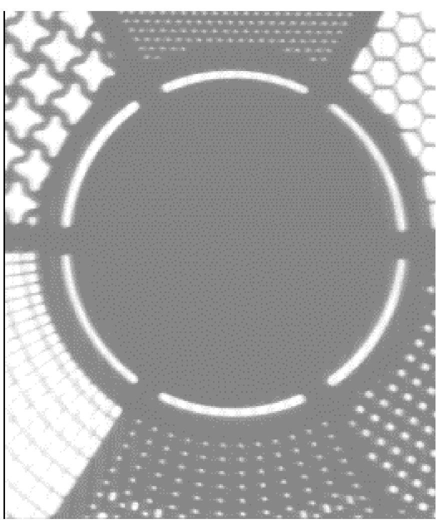

Figure 7: Image of a 2 mm thick collimator pattern taken using a 1.5 mm thick hybrid silicon pixel detector with a 49 kV x-ray generator at 40 mA. The smallest holes shown at the top are 0.15 mm in diameter.

All the images shown here are raw data with simple background correction. The silicon pixel detectors show very small background variations while the CdZnTe pixel detector shows a small amount of visible band structure in the image. The non-uniformity of the CdZnTe detector material and collecting holes instead of electrons are most likely causes of such variances. Since these are digital images, it is possible to calibrate and correct the data and improve the uniformity of the image quality. Extensive research is being carried out to improve CdZnTe material quality significantly in the near future.

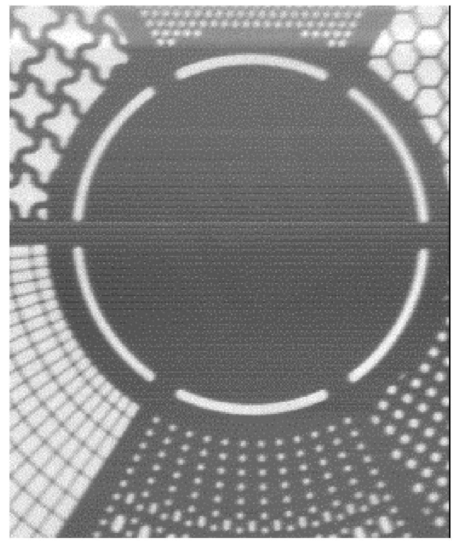

Figure 8: Image of the same collimator patterns taken using a 2-mm thick CdZnTe hybrid pixel detector. The smallest holes shown at the top are 0.15 mm in diameter.

IV. DISCUSSION

The test data for the silicon pixel detector hybrids show excellent images, which can be applied in medical and industrial applications.

Figure 8 shows the first image taken from a 2-mm thick CdZnTe detector. It is well known that the mobility-life time product for holes generated in this material is on the order of 10^{-5} cm^2/V [7], and the total distance that holes can travel under a bias voltage of 600 V is about 0.3 mm. This means that the charge generated by the x-ray will not be fully collected in a detector thicker than this. In contrast, the electron mobility-life time product in CdZnTe is about 2 orders of magnitude larger than that of the holes. For historical reasons, the present version of the readout chip was designed to collect holes, and this would limit the usefulness of the detector for certain applications. However, results from the hole-collecting readout chip give us valuable information about possible performance of future electron-collecting readout chips, which are expected to produce significantly improved images. For mammography applications, the x-ray energy is about 20 keV, and a thickness of about 0.3 mm of CdZnTe would be adequate to stop the x-rays. We are planning to trim the detector thickness down to about 0.3 mm and perform further tests.

We did see some sections which did not yield any output for the CdZnTe detectors, probably due to polarization and charge trapping, as discussed earlier. From Figure 8, we see that the overall image is not affected by these bad sections, indicating that the bad sections were just averaged out from the good ones. This again proves our TDI design concept that the system is immune to a few bad detector sections.

V. Summary and Future Work

Preliminary results show that the concept of direct conversion solid state detectors is feasible, with possible superior performance than that of the conventional x-ray films or scintillator-based systems. An electron-collecting readout chip is desirable for use with the CdZnTe detectors. The system demonstrated above provides new approaches in developing high resolution, high DQE and high dynamic range digital radiographic instruments for medical or industrial applications.

For our future work, we will trim down the 2-mm thick CdZnTe detectors to about 0.3-mm and characterize their performance for mammography application. Quantitative DQE measurements of these detectors are underway. We are also planning to redesign the readout chip where an electron collection version of the readout chip can provide wider energy range and improved performance.

VI. References

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