A Low-Dose High Contrast Digital Mammography System (DigiMAM)[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 also 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 photons and light photons to electron-hole pairs 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, is designed to have linear response to x-rays, and can support a dynamic range of 14 bits. The detector arrays are lightweight and can be packaged compactly, facilitating their incorporation into a digital mammography system or into other clinical radiology and industrial imaging applications where high sensitivity, excellent contrast, high resolution, and high dynamic range are required.

In this paper, we report on tests of the first full size prototype readout ASIC chips hybridized with either Silicon or CdZnTe detector arrays and present initial results and images.

I. Introduction

Digital mammography offers the potential of improved image quality and the possibility of increased performance of detection of 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 must convert the x-ray pattern transmitted by the breast into an electronic signal 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 detector area.

Most current detector approaches for digital mammography employ a phosphor x-ray detector which, in response to x-ray absorption, produces 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 phosphor type detectors by using direct conversion solid state detectors. In these detectors the x-ray photon directly creates electron-hole pairs without producing the intermediate light photons. Such direct conversion detectors require custom-made readout electronics. Since the pixel sizes are extremely small (50 x 50 micron) it is a major challenge to obtain coupling between all of the detector elements and the readout electronics. We have employed an innovative technology, indium bump bonding, to achieve coupling by hybridizing a detector layer to a readout layer. Since the detector and the readout chip are manufactured separately they can be optimized independently to achieve maximum quality.

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

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

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

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

II. System Design

A. Detector Design

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

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Silicon has the advantage of easy processing. The present technology in processing silicon wafers is mature and the material is cheap compared to CdZnTe. Since silicon has an atomic number of 14, it has a low x-ray attenuation coefficient and is expected to have a DQE of about 60% for a 1 mm silicon detector for 20 keV x-rays. It needs, 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 contributes to high signal to noise ratio (SNR). Both 1 mm and 1.5 mm thick silicon detectors have been tested using the prototype system.

CdZnTe has several properties that make it potentially useful for digital mammography and other industrial applications. It 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 [2]. There are a few drawbacks associated with this new detector material. The processing technology is not mature yet and the detectors have charge trapping centers for crystals grown using the high-pressure Bridgman technique [3]. Another disadvantage of this material is that its hole mobility-lifetime product is about two orders of magnitude smaller than that for the electron. Therefore, it is important to collect the electrons rather than the holes [3][4]. The cost of this material is also high at present due to low yield. For the prototype system a 2 mm thick CdZnTe detector was fabricated and hybridized to the existing hole-collecting MARY readout chip. It is anticipated that some charge collection inefficiency is inevitable for this prototype system. An electron-collecting version of the MARY chip is in preparation.

B. MARY Readout Chip

The MARY chip employs time delay integration (TDI) in which there are 192 TDI stages. There are 2 versions of the MARY chip: a large MARY chip in which there are 384 TDI columns and the small MARY chip in which there are 128 TDI columns (Figure 1).

Because of the high sensitivity of the detector materials selected, the large amount of electron-hole pairs can easily saturate the chip's well capacity. To handle the large charge overflow, the 192 TDI pixels are divided into 24 sections (i.e. 8 TDI stages per section) with a readout buffer attached to each. This design has the advantage that sections with defective pixels will not affect the performance of other sections. The dynamic range is designed to be 14 bits. The maximum well capacity is 400 million electrons and the readout noise is about 3,000 electrons for each output section. The chip is designed for fan beam scanning applications, and for this purpose multiple units can be abutted in a staggered linear array formation up to any practical length.

The MARY chip is designed to handle slot-scan x-rays where the charge transfer speed is about 1 ms per step. The charge transfer efficiency was measured to be 0.998 per transfer step.

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.

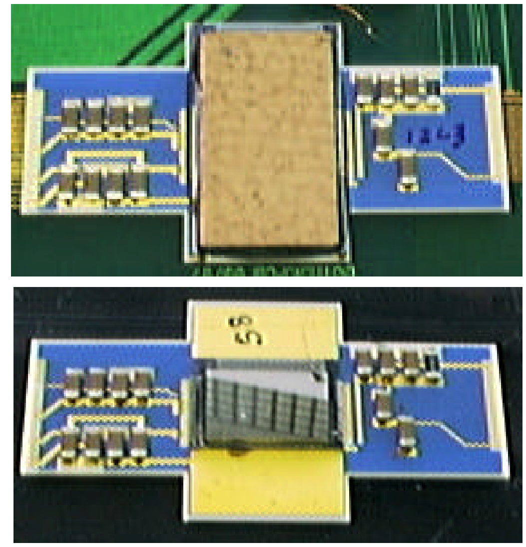

Figure 1: A large MARY chip (Top), hybridized with a 2 mm thick CdZnTe detector, and a small MARY chip (Bottom), hybridized with a 1 mm thick silicon detector, mounted on ceramic chip carriers.

III. TEST RESULTS

The detector 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. Figure 2 shows such images where the 9 lp/mm patterns are clearly visible (the Nyquist limit for 50 micron spatial resolution is 10 lp/mm). The detector is a 1 mm thick silicon PIN diode array where holes are collected in the MARY readout chip.

Figure 3 is a MTF plot measured from the same detector hybrid where MTF functions from both scan and slot directions are plotted.

Figure 4 is a finger phantom image plotted in log scale, again using the 1 mm thick silicon detector hybrid. Since a small MARY chip was used, several parallel scans were taken and the images were spliced together in software to produce the larger image.

A 1.5 mm silicon detector hybrid was also tested. 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 a jaw with teeth.

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

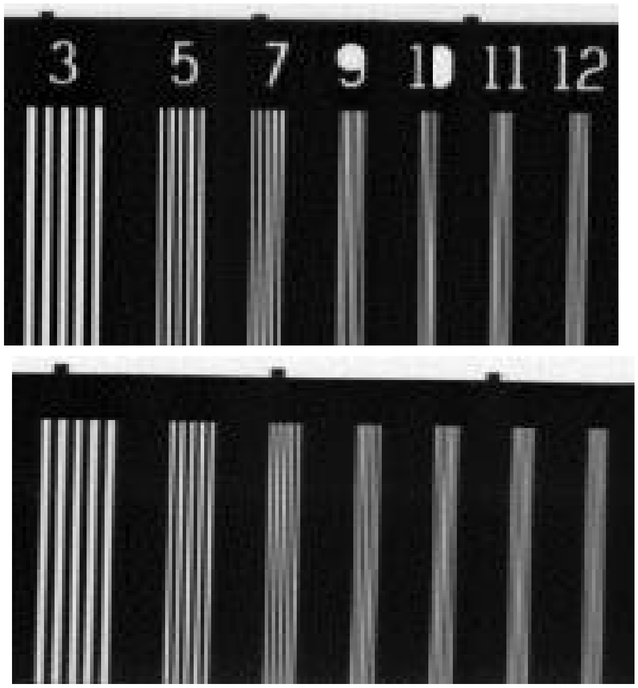

Figure 2: Bar pattern x-ray images using a 1 mm thick silicon pixel detector hybrid. The 9 lp/mm pattern is resolved clearly in the top picture (scan is in the non-TDI direction). In the bottom picture the scan is in the TDI (horizontal) direction.

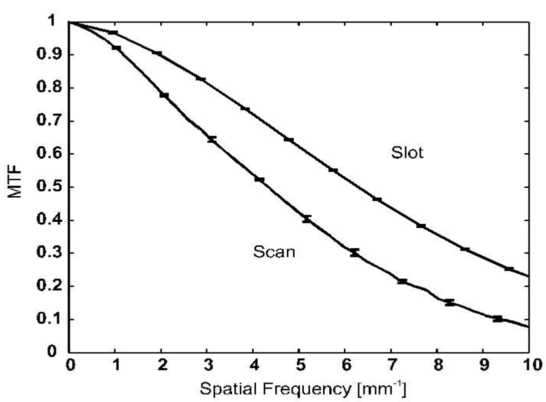

Figure 3: MTF plots of a 1 mm thick silicon pixel detector. The data in both TDI (scan) and non-TDI (slot) directions are shown with standard error with 5 repeated measurements.

A 2 mm CdZnTe pixel detector hybrid (the first prototype which has just been fabricated) was also tested under a different machine where the x-ray generator was set to 160 kV and the tube current was 2 mA. The focal spot of the tube was 0.2 mm in diameter and about 50 cm away from the detector. The detector bias was set at 600 V. Figure 8 is a picture of the same collimator pattern as 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.

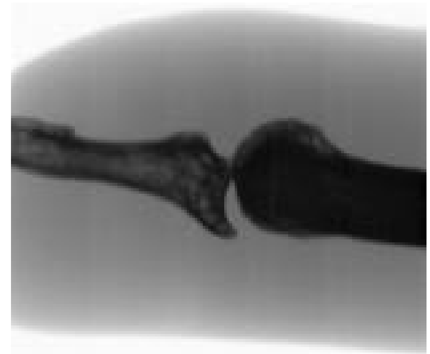

Figure 4: X-ray image of a finger phantom using a 1 mm thick hybrid silicon pixel detector.

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

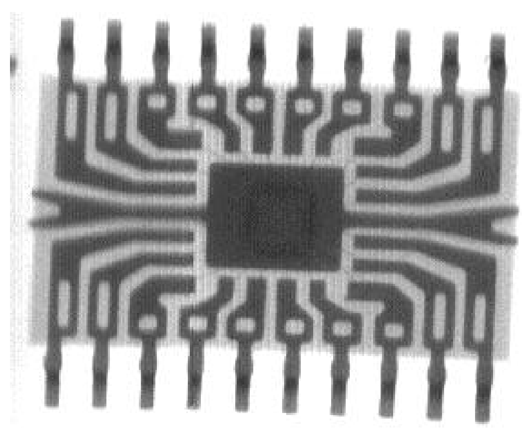

Figure 6: Image of an IC placed at a slight angle to the x-ray scanning direction using a 1.5 mm thick silicon pixel detector. The 0.025 mm diameter wire bonds are clearly visible.

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.

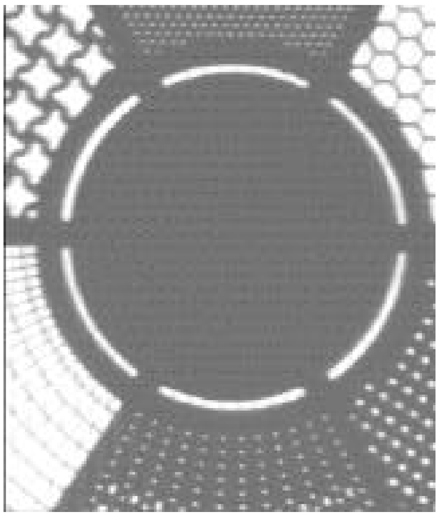

Figure 7: Image of a 2 mm thick collimator pattern taken using a 1.5 mm thick hybrid silicon pixel detector with a 49 kV x-ray generator at 40 mA. The smallest holes shown at the top are 0.15 mm in diameter.

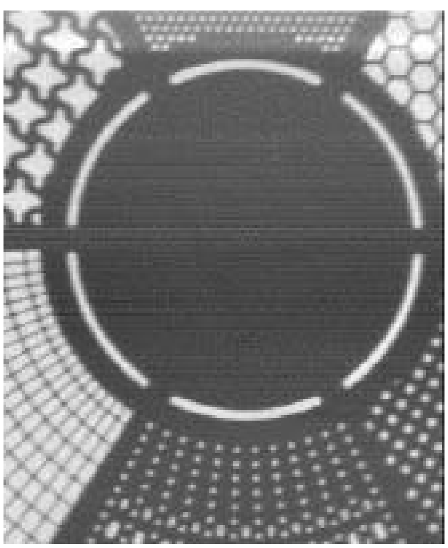

Figure 8: Image of the same collimator pattern taken using a 2 mm thick CdZnTe hybrid pixel detector. The smallest holes shown at the top are 0.15 mm in diameter.

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 [5], 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 MARY chip was designed to collect holes instead of electrons, and this would limit the usefulness of the detector for certain applications. However, results from the hole-collecting MARY chip give us valuable information about the performance of future electron-collecting MARY 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 to about 0.3 mm and perform further tests.

V. SUMMARY AND FUTURE WORK

Preliminary results show that the concept of direct conversion solid state detectors is feasible, with possible superior performance than the conventional x-ray films or scintillator-based systems. An electron-collecting MARY 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 thickness 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 MARY chip can provide wider energy range and improved performance.

VI. REFERENCES

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