Mid-IR photon counting array using HgCdTe APDs and the Medipix2 ROIC

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ABSTRACT

We present the design of an infrared (IR) photon counting array consisting of an array of mid IR HgCdTe APDs read out with a CMOS application specific integrated circuit (ASIC) developed for x-ray imaging called the Medipix2. The Medipix2 is an array of 256x256 pixels, each of which amplifies and counts pulse events. When combined with an APD whose gain is high enough, the Medipix2 will integrate these detected photons, and the binary readout of the counters will be fast (~1kHz framerate) and noiseless. Initial feasibility tests of this concept using individual APDs from DRS technologies Inc. wirebonded to Medipix input pads are discussed.

Keywords: IR detectors, adaptive optics, wavefront sensors, Medipix, photon counting, IR APD arrays

1. INTRODUCTION

Low readout noise in imaging detectors in the infrared has been a goal for decades. In applications where the background fluence per frame is low, the signal to noise ratio of the image is dominated by the readout noise. Examples of this low background regime include IR wavefront sensors for adaptive optics (~kHz frame rates), narrow band imaging, spectrophotometry, interferometry and space-based imaging.

Most array-based sensors of optical and infrared radiation integrate the charge generated by the input photon flux in a conversion layer at the pixel. This charge is then transferred to a charge-to-voltage analog amplifier and this voltage is converted to a digital number by a downstream ADC. Unfortunately, there is a noise associated with the charge to voltage amplification and conversion. This noise increases with increasing bandwidth, so faster readout (frame) rates suffer a readout noise penalty. Also, charge integrating detectors will integrate internally generated leakage current, so infrared detectors with smaller bandgaps must be operated cryogenically and read out fast to minimize this additional charge and its associated noise.

The detector of choice for most array-based optical detectors are silicon charge coupled devices (CCDs), which have high QE in the optical. Given the standard CCD readout schemes of clocking the detected charge through a finite number of on-chip amplifiers, there is a direct tradeoff between the number of pixels readout per second and the readout noise per pixel. Recent efforts to improve CCD readouts include: completely parallel readouts with one amplifier per column1 which allows a slower clock rate for a given frame rate; gain stages in the serial readout structure that amplify the charge signal before the signal is sensed by the readout amplifier (L3CCD by E2V technologies2) which reduces the effective readout noise; and clever geometrical pixel layout structures that match the multiple regions of interest, thereby simultaneously reducing the number of pixels and increasing the number of parallel readout amplifiers.3

In the near and mid-infrared, the photon is converted to a charge in an absorbing material (e.g. HgCdTe, InGaAs) with a smaller bandgap and higher IR quantum efficiency than silicon. This absorber is usually coupled to a silicon CMOS read out integrated circuit (ROIC) which integrates the collected charge and can use many of the same readout strategies discussed above to minimize the readout noise, e.g. parallel fast readouts to reduce the dark current contribution, multiple sampling of the charge signal, and smaller region of interest sampling. Yet, very low noise readout at high frame rates in the near and mid-IR for large array detectors still do not exist. The NIRSpec focal plane array for the James Webb Space telescope has two 2048x2048 HgCdTe sensors based on the Rockwell HAWAII-2RG chip with a readout noise of 6 e− rms per 1008 second exposure4.

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1.1 Photon counting detectors

Alternatives to charge integrating arrays are photon counting detectors. “Photon counting” detectors are so named because photon events are detected and counted individually. A photon counting detector can detect a single photon without ambiguity, enabling the downstream electronics to increment (by one) some sort of counter in a histogram, either spatial, spectral or temporal. In the UV/optical regime, photon interactions with matter usually result in a single photoelectron that is difficult to detect directly with simple linear amplifiers. If this single photoelectron can be amplified before external measurement, such that the resulting number of electrons significantly exceeds the downstream amplifier noise, then the amplifier output can be used to count the photons directly using a discriminator circuit. The detectors are considered “noiseless” in that the readout of N events (whole number) implies that N events were detected. This is not to say that the next measurement of the same phenomena will be N events, that is governed by the statistics of the photon arrival (“photon noise”). It is the readout noise of photon counters that is zero. [Often, the background event rate in photon counting detectors due to cosmic rays, scattered light and thermionic emission are referred to as “noise” or “dark noise” but the detection and counting of those events are real measurements and the term “noiseless” hereafter in this paper refers strictly to the readout noise.]

Photon counting is common in x and gamma ray imaging, where the individual photon energy is great enough to generate many electrons in the absorber itself and can exceed the readout noise of the amplifier. In the UV and optical bands, electron amplification is used to convert single photoelectrons into larger electron pulses. Examples include photomultipliers with their dynode structures or microchannel plates (MCPs) that preserve the location of the incident photon while it is readout through patterned anodes or phosphor screens (intensifiers). Depending on the readout technique, electron gains of $10^4$ to $10^5$ might be required to detect individual photons well above the noise of the electronic amplification.

Photon counting imaging detectors not only count the event but also provide the x,y position of each event and sometimes the time, $t$, of each event. Examples in the UV and optical include avalanche photodiode (APDs) arrays and imaging MCP detectors. Silicon based APDs are fast and potentially have the high QE of silicon in the optical and near IR but are not yet incorporated in large arrays. Imaging microchannel plate (MCP) detectors can have large area (100 x 100 mm), high spatial resolution (20 µm FWHM), low background dark count, and event timing resolution less than 1 nanosec. Their QE is determined by the characteristics of the photocathode material that absorbs the initial photon and releases the photoelectron. The best photocathodes in the optical/IR are based on GaAs (cutoff $\lambda < 800$ nm) and InGaAs ($\lambda < 1600$ nm), but the QE tends to be limited to $<40\%$.

A penalty for photon counting is the associated deadtime per event when either the input transducer (photon to photoelectron to amplified pulse) or downstream counting electronics are busy and cannot process another event. The deadtime per event varies with detector, and can range from 10ns to 10µs. For random and uncorrelated input flux (e.g. starlight) a deadtime of a microsecond would correspond to a 10% deadtime for a 100kHz average input rate, i.e. 10% of the input photons would not be detected. This loss can be completely characterized, calibrated and corrected, but it is equivalent to a rate dependent QE loss. Therefore to operate at high detector input flux, many independent parallel channels must be used. Counting each photon allows arbitrary integration times, which can be very fast, dependent only on the speed and dynamic range of the counter. However, for high rate parallel counter systems, the speed of the counter readout must also be factored into the integration time. The time it would take to readout all the counters in a 2D array would represent an effective minimum time frame.

1.2 Infrared avalanche photodiodes

Avalanche photodiodes (APDs) operate near the breakdown voltage of the diode (linear) or slightly above the breakdown voltage (Geiger mode) allowing a single photon can initiate an avalanche of charge carriers. If the gain, $G$, of this avalanche/amplification process is constant and independent of the number of input photoelectrons, then the internal amplifier is said to be proportional (or linear), i.e. the output amplifier pulse amplitude is proportional to the input number of photons. If the gain is not constant from pulse to pulse, the output signal has an excess noise due to the amplification process itself.

If the gain mechanism can saturate, such that any size input (except zero) results in the same output amplitude, the amplifier is said to be in “Geiger” mode, where all information of the input number of events for that pulse is lost except that it was equal to or greater than 1 event. There is no “excess noise” because the amplitude of the event is not used, just the existence of the event itself. However, if the Geiger mode detector can recover fast enough such that the probability
of 2 events occurring in this time is very low, then simply counting the Geiger mode events in an integration time gives the detected input flux. This mode is normally used in photon counting as the gain is higher and the noise in the amplification process does not matter as long as the event is above threshold.

Infrared APDs were initially developed in Germanium and InGaAs for fiber optic communications in the linear regime at room temperature, but to avoid thermally generated events, they had to be cooled below room temperature (220K to 90K). Higher gain single photon avalanche diodes (SPADs) are biased above the breakdown voltage and operate in the Geiger mode, but infrared sensitive InGaAs SPADs have proven difficult to operate in non-gated operation due to afterpulsing and poor depletion at cold temperatures. Linear mode APDs have a much lower gain than Geiger mode SPADs, and would prove difficult to trigger a fast counter circuit, especially with excess noise factors due to the amplification process. However, a recent development of a unique infrared APD structure by DRS Technologies Inc. has achieved a gain > 1000 in the linear regime with an excess noise factor of ~ 1.0 based on the “high density vertically integrated photodiode” (HDVIP) architecture. These APDs have also been fabricated into arrays and if a ROIC existed that could count these very small pulses, then an imaging infrared photon counter would be born.

1.3 Medipix2 CMOS ROIC

Such a ROIC exists, but it was developed for a completely different application: x and gamma-ray imaging diode array readouts. The Medipix2 chip is a pixel detector readout chip consisting of 256 x 256 identical elements, each working in a single photon counting mode (Fig. 1). It was developed by the Microelectronics Group at CERN for the Medipix Collaboration. The low noise of the input amplifier (≤ 110 e−) allows the setting of a very low discriminator level (≤900 e−) to reject amplifier "events", yet it can potentially detect all events from an APD with a gain of 1000 and a small excess noise factor of 1. Such an excess noise factor would imply that photon events from the APD would have an rms dispersion of 31 electrons at a gain of 1000.

Our group at the Space Sciences Laboratory, Univ. of California, Berkeley, is a member of the Medipix collaboration and has been using the Medipix2 ROIC to integrate counts from imaging MCP detectors. Normally, for x-ray imaging, a silicon diode array is bump bonded to the Medipix2 chip for imaging of individual x-ray events. Our idea was to extend the Medipix2 attributes (kHz frame rate, photon counting, electronic shutter, etc. see below) into the UV/optical by using an MCP as an amplifier of single photoelectrons from a photocathode in a vacuum tube. We were funded in this endeavor by the National Science Foundation through the National Optical Astronomy Observatory to develop a noiseless optical detector that could operate at kilohertz frame rates for future wavefront sensors.

Extending this idea into the infrared, where most of the adaptive optics observations are performed, we needed a transducer that could convert a single IR photon into an electron pulse of > 1000 e−, be fabricated into a large array and could be coupled to our Medipix2 pixel sizes of 55 microns square. We also needed to show that the Medipix2 chip would operate at the cryogenic temperatures (~77K) required by the IR APD.

2. HDVIP INFRARED APDs

Developed by DRS Technologies, Inc., the APDs are made in short, mid and long wavelength cutoff (\( \lambda_{c} \)) infrared HgCdTe. DRS Technologies utilize a cylindrical "p-around-n", front side illuminated, n+/n/p geometry that favors electron injection into the gain region (Fig. 2). The reason for the good match to the Medipix ASIC is that the cylindrical
APD structure around a wet-etch via can directly couple the output signals to the Medipix input pads underneath the APD array. These APD devices are characterized by a uniform, exponential, gain vs. applied bias voltage. Gains of greater than 1000 have been measured in $\lambda_c = 4.3 \, \mu m$ Hg$_{0.7}$Cd$_{0.3}$Te (Fig. 3). At 80ºK, $\lambda_c = 4.3 \, \mu m$ devices show excess noise factors of close to unity out to gains of 1000$^9$.

The noiseless amplification process of this type of APD is important for its potential use with the Medipix chip. The input amps of the Medipix2 have an rms input noise of ~110 electrons. If a minimum threshold for counting events is set at ~9 times the rms noise level (i.e. 900 electrons), then the APD must have a gain above this level. There will be a distribution of event sizes about the gain level, and for an amplifier with gain ~ 1000 and zero excess noise, this distribution will have an rms value of ~32 electrons, so all events will be above threshold and therefore counted. The APD does not need to be operated in the Geiger mode to be counted. This can be compared to MCPs, where the low gain amplification noise is high, resulting in an exponential output pulse distribution, and loss of events below the Medipix threshold. (For MCPs, this can be overcome by simply increasing the gain of the MCPs by raising the high voltage).

Preliminary discussions with scientists and engineers at DRS Technologies Inc. have led us to believe that both technologies are a good fit. To get the required gain by the Medipix2 chip (> 1000) requires a smaller bandgap ($\lambda_c \sim 4.3 \, \mu m$) and therefore we would expect to operate at 77ºK. Though designed as a room temperature device, the Medipix2 is expected to be able to function at 77ºK as it is a CMOS device. In fact, there is a possibility it might run with less amplifier noise and faster.

3. THE MEDIPIX2 ROIC

The Medipix2 chip (version Mpix2MXR20) is a pixel detector readout chip consisting of 256 x 256 identical elements, each working in a single photon counting mode. Each pixel consists of a preamplifier, a discriminator, and a 14 bit pseudo-random counter. The counter logic, based on a shift register, also behaves as the input/output register for the pixel. Each cell also
has an 8 bit configuration register which allows masking, testing and 3-bit individual threshold adjust for the discriminator (Fig. 4). It was designed and manufactured in a six-metal 0.25 µm CMOS technology. Each 55 x 55 µm pixel contains 530 transistors. The total active area of the chip is 1.98 cm² and is 3-side abuttable to support larger arrays. Fig. 5 shows the electrical schematic of the Medipix2 layout and how the shift registers relate to the parallel readout.

The input referred noise of the Medipix2 cell has been measured to be ~110 e⁻rms and the thresholds can be adjusted to a consistency of ~95 e⁻rms. Note that this noise is in the detection circuit and all we require of the circuit is a minimum of false events. For the minimum threshold setting of 900 e⁻, a false trigger would be an 8 sigma noise fluctuation.

The Medipix2 has two working modes depending on the Shutter input. When the Shutter signal is low, the pixel is in acquisition mode and the discriminator clocks the counter. When the Shutter signal is high, acquisition stops and an external clock is used to shift the data from pixel to pixel out along the columns (Fig. 5). Again note that this is a noiseless process, digital bits are being transferred, not charge like in the case of a CCD. For the Medipix2, this has been measured at 100 MHz and modeled up to 150 MHz so to read out the whole array will take 284 µs. Note that the current design does not allow data collection while reading out, so the 266 µs is a fixed deadtime per frame readout. At a kHz rate, the sensor is active 73.4% of the time which translates into less sensitivity.

The Medipix consortium is now embarking on the next generation ASIC using 130nm CMOS technology (“Medipix 3”). Halving the feature size will allow the addition of extra counter buffers in each pixel so the ASIC can integrate while being read out. The front end amplifiers will have higher bandwidth and therefore less deadtime. These changes coupled with faster serial and parallel readouts should result in frame rates up to 10 kHz. The chip will also be designed to be radiation hard. The 55 µm pixel size will probably stay the same, to accommodate the x-ray and gamma ray imaging applications. Initial analog tests of an 8x8 prototype test structure look promising and delivery of the full 256x256 array is expected in 2009.
4. HDVIP IR APD - MEDIPIX2 HYBRID

These two technologies, the infrared APD and the Medipix2 ROIC were initially designed for different applications and certainly not designed with the other in mind. However, there are certain aspects of each that lend themselves to a hybridization into a low noise IR imaging detector. First, the gain of the HDVIP APD can exceed the minimum detectable input signal of the Medipix2 chip (~ 900e−) by using the $\lambda = 4.3 \, \mu m$ cutoff formulation of the HgCdTe where gains can exceed 1000. Second, the pixel size of the Medipix2 (55 \, \mu m) is close to existing array sizes of the APD (8x8 array of 64 \, \mu m pixels). Third, the excess noise factor of ~1 for these APDs indicate that all photon events will be above threshold if the gain is sufficiently high. This is in contrast to the exponential gain distributions of single MCPs where the gain must be very high to get the smallest output signals above threshold. Fourth, the technique for making the HDVIP APDs by wet etching a via through the planar HgCdTe allows the signal input to be directly coupled to the Medipix2 input pads without chip flipping or bump bonding. And finally, the Medipix2 can accept either electron or holes as inputs, and can accommodate leakage currents up to many nanoamps per pixel, enough for these APDs at high gain and 77K.

In contrast to the fortuitous similarities in characteristics and specifications, there are aspects of each technology that might "doom the marriage". The Medipix2 chip was designed for room temperature operation, not 77K. There is no provision to separately bias each APD in an array, which places a uniformity requirement on the APD array. The Medipix2 input circuit has no protection circuit in place, so if an APD fails shorted, the whole chip might be destroyed. Also, the Medipix2, though designed as a "low power" chip, operates at 0.5 watt in quiescence, which is rather high for a cryogenic detector.

4.1 Medipix2 cold test

Before proceeding with any APD hybridization with a Medipix2, we tested the characteristics of the Medipix2 at cryogenic temperatures down to 77K. We used a Medipix2 chip that had been epoxied onto an alumina header that we used for our imaging vacuum tube project\(^{15}\). This chip had already been cycled to 350C (non-operating) to test its ability to survive the high temperatures of vacuum tube processing and it survived with little change in its characteristics. The chip-header was mounted onto a printed circuit board with a 68 pin readout cable on one side and a 10 cm copper finger on the other. This assembly was dangled over liquid nitrogen in a glass dewar, and slowly lowered while its readout and noise characteristics were measured. Being a bare readout chip, we could not measure input signals other than test pulses. But we could confirm that the digital readout operated and that the digital to analog converters (DACs) used to adjust the operating point of the circuits produce voltages and currents in range to run the chip. We could also scan the input analog threshold through the amplifier output baseline and thereby measure a noise distribution.

![Fig. 6. Medipix2 DAC output scans at 5 different temperatures. The "GND" signal is a bias voltage used to set the DC bias of the input node. The "PREAMP" signal is a bias current in the front-end amplifier. The feedback voltage monitor for the current DACs is the transistor gate voltage, which does not directly reflect the full range of currents available. Though we see some odd effects at certain DAC input values and low temperatures (GND > 150 units), the stability over a > 200K range attests to a robust design.](image-url)
The test was successful in that the Medipix2 worked at 77K, though not the same as at room temperature. The latest version of the Medipix2 chip (MPX2MXR) has many of its front end circuits inherently temperature compensated. Fig. 6 show a scan of DAC analog output values at various chip temperatures. Fig. 7 shows a threshold scan through the noise at room temperature and at 77K, showing that the 77K results are not optimal. However, re-optimizing the various DAC settings resulted in the threshold scan in Fig 7. This scan indicates that either the noise increased at low temperature, or that the gain increased and that the units of the measurement were different. This would be expected as the discriminator input amplifier has an input NMOS transistor whose transconductance should be sensitive to temperature and this is not compensated in the design.

4.2 HVDIP IR APD - Medipix feasibility tests

Fabricating a full HgCdTe APD array onto a Medipix2 ROIC would be a bit premature without some feasibility tests, given the possible showstoppers mentioned above and the significant costs of design, mask sets, and fabrication runs. We decided to test whether we could detect IR photons by directly coupling the APD outputs to Medipix input pads using wirebond, cycling the crude hybrid to 77K, and illuminating it with mid-IR photons. This is a non-optimal test in that the amplifier noise will increase substantially due to the increased capacitive loading of the wirebonds and the fan-out traces on existing APD array carriers. However, we can overcome this noise by using a pulsed infrared photodiode where we can control the intensity and width of the pulse. Such a pulse will have many photons arrive at the APD during the Medipix2 pulse shaping interval (~300ns) and therefore counted as a single (but larger) pulse, well above the threshold set above the noise floor. For example, if we adjust the LED to give 16 photons per pulse (in a time interval of <10 ns) with an APD gain of 1000, we would...
expect pulse inputs of 16,000 e\(^{-}\) with an rms distribution of \(\sqrt{16} \times 1000\) or 4000 e\(^{-}\). This error in signal would dominate any variance in APD amplification or Medipix amplifier noise.

An 8x8 HgCdTe HDVIP array (MWIR \(\lambda_c \sim 4.3\) \(\mu m\)) with 64 \(\mu m\) pixels was previously mounted to a fan-out board. This fan-out was mounted next to the Medipix chip (Fig. 8) and shares the same 100 pin Leadless Chip Carrier (LCC). Eight diodes from the APD fan-out chip were wire bonded to inputs on the Medipix Chip. We used pixels near the edge of the Medipix2 to keep the wirebond lengths short.

The I/O pads that are required to run the Medipix chip are bonded to the LCC. The LCC is mechanically mounted in a combined Lakeshore chip carrier/personality card (Fig. 9). The personality card has available 100 leads that can be configured for a given test set-up as well as a ground plane; in this case filter capacitors are mounted on the card and attached to all of the supplies and the ground plane. Each of the 100 pins from the personality card is electrically carried to the outside of the dewar, where they terminate in four, fifty pin connectors: 25 of the pins in each connector are for signals and 25 from each connector are ground connections. A custom cable was made to bring the signals from the dewar to the Medipix USB interface box\(^\text{16}\) and then on to a PC via a USB cable. The Pixelman software\(^\text{17}\) will allow us to adjust the DACs to optimize performance of the eight pixels in terms of noise and to externally stimulate the Medipix2 pixels to independently measure the gain.

![Fig. 9. The Medipix2-APD test hybrid mounted on the LCC that in turn is mounted in the dewar. The personality card where bias resistors and capacitors are mounted is directly underneath and surrounding the LCC.](image)
The eight HgCdTe APDs can be stimulated with an EG&G 3.5 µm cutoff IR LED, which is mounted in close proximity to the APD fan-out. The housing that holds the LED (Fig. 10) completely covers the LCC and is blackened to prevent light leaks. By stimulating the APDs with different pulse sizes and running threshold scans, we can generate a photon transfer curve where the log of the average signal level is plotted against the rms of the signal amplitude. The slope of this curve should be 0.5 in the higher flux limit and plateau to the Medipix2 amplifier noise. The extrapolation of the high flux line to the log(signal) = 0 axis gives the gain of the system.

4.3 Results (or lack thereof)

At the deadline for submission of this paper, the APD-Medipix hybrid was wired up and ready to turn on at room temperature. After room temperature optimization of the Medipix2 readout, we will lower the temperature to 77 K and re-optimize the voltage and current biases of the Medipix2. We will then bias the APDs at low (~100) gain and proceed to characterize the pixels at high flux levels before progressing to higher gain (~1000), in case there is breakdown of one of the APDs that might destroy the Medipix2.

If these tests prove the feasibility of MWIR photon counting, we will seek funding to proceed with HgCdTe arrays fabricated directly onto the Medipix2 ROIC itself, either at the die or wafer level to demonstrate imaging and photon counting down to the single MWIR photon level.

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