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Photon Counting Energy Dispersive Detector Arrays for X-ray Imaging

Jan S. Iwaczyk¹, Einar Nygård², Oded Meirav³, Jerry Arenson³, William C. Barber¹, Neal E. Hartsough¹, Nail Malakhov², and Jan C. Wessel²

¹ DxRay Inc., Northridge, CA 91324 USA

² DxRay, Inc. and Interon AS, Asken, Norway

³ GE Global Research, Haifa, Israel

Abstract

The development of an innovative detector technology for photon-counting in X-ray imaging is reported. This new generation of detectors, based on pixellated cadmium telluride (CdTe) and cadmium zinc telluride (CZT) detector arrays electrically connected to application specific integrated circuits (ASICs) for readout, will produce fast and highly efficient photon-counting and energy-dispersive X-ray imaging. There are a number of applications that can greatly benefit from these novel imagers including mammography, planar radiography, and computed tomography (CT). Systems based on this new detector technology can provide compositional analysis of tissue through spectroscopic X-ray imaging, significantly improve overall image quality, and may significantly reduce X-ray dose to the patient. A very high X-ray flux is utilized in many of these applications. For example, CT scanners can produce ~ 100 Mphotons/mm²/s in the unattenuated beam. High flux is required in order to collect sufficient photon statistics in the measurement of the transmitted flux (attenuated beam) during the very short time frame of a CT scan. This high count rate combined with a need for high detection efficiency requires the development of detector structures that can provide a response signal much faster than the transit time of carriers over the whole detector thickness. We have developed CdTe and CZT detector array structures which are 3 mm thick with 16×16 pixels and a 1 mm pixel pitch. These structures, in the two different implementations presented here, utilize either a small pixel effect or a drift phenomenon. An energy resolution of 4.75% at 122 keV has been obtained with a 30 ns peaking time using discrete electronics and a ⁵⁷Co source. An output rate of 6×10^6 counts per second per individual pixel has been obtained with our ASIC readout electronics and a clinical CT X-ray tube. Additionally, the first clinical CT images, taken with several of our prototype photon-counting and energy-dispersive detector modules, are shown.

Index Terms

CdTe; CdZnTe; photon counting computed tomography; X-ray detectors

I. Introduction

There is a need for the development of novel detector technologies for photon-counting in X-ray imaging. The applications that can greatly benefit from these novel imagers include mammography, planar radiography, and computed tomography (CT). In order to improve contrast at reduced dose, photon-counting detectors have been considered to develop

alternatives to conventional X-ray detectors used in radiography and CT [1]–[3]. Photon-counting X-ray detectors applied to mammography and radiography have shown a dose reduction of 40% to 400% while maintaining sufficient contrast for these applications [4], [5]. Simulations have shown that photon-counting along with optimal energy weighting can increase dose efficiency up to 40% for digital mammography as compared to a conventional integrating system [6]. One group has reported that the lesion detection signal to noise ratio is significantly affected by the energy dependence of the detector's quantum efficiency and the varying contrast carried by different energies in the beam spectrum [4]. Recent studies have demonstrated that weighting photons by a factor of E^{-3} can lead to an improved SNR compared to simple photon-counting or energy weighting (as in intensity-integrating detectors) [7]. In another simulation study it was demonstrated that a photon-counting detector can reduce beam hardening and improve lesion signal to noise ratio compared to conventional detectors; optimally weighting the photons further improved the signal to noise ratio but resulted in greater beam hardening due to the greater weight given to lower energy photons [8]. Photon-counting can provide similar improvements to CT provided that sufficient performance detectors can be developed.

There are number of detector structures that have been proposed over the years to improve performance of CdTe and CZT detectors including for example co-planar grids contact configuration [9]. These types of detector structures are specifically intended for large volumetric detectors for gamma ray spectroscopy. The X-ray imaging detectors presented here are based on highly pixellated cadmium telluride (CdTe) and cadmium zinc telluride (CZT) detector arrays electrically connected to application specific integrated circuits (ASICs). The single photon-counting readout is accomplished within the ASIC and the close integration of the sensor material and readout can produce fast and highly efficient photon-counting and energy-dispersive X-ray detectors. Systems based on this new detector technology can provide compositional analysis of tissue through spectroscopic X-ray imaging, significantly improve overall image quality, and may significantly reduce X-ray dose to the patient. Very high X-ray flux are utilized in many of these applications [10], [11]. For example, CT scanners require ~ 100 Mphotons/mm²/s in order to collect sufficient photon statistics during the very short time frame of the measurement. To develop a photon-counting X-ray imaging detectors for CT, the high count rate combined with a need for high detection efficiency requires the development of detector structures that can provide formation of the response signal much faster than the transit time of carriers over the whole detector thickness. We, as well as others, have investigated a number of strategies pursuing the development of detector structures and electronic readouts that are capable of a maximum output count rate sufficient for CT imaging. The investigated detector array configurations include the following attributes which we make use of to increase the maximum count rate: a) small pixel effect, b) parallel drift structures, c) multiple layers, and d) a combination of above. There are merits and limitations inherent with each of the four approaches. Detector electrode configurations creating the so called "small pixel effect" [12] are very effective in reducing the duration of the fast portion of an induced signal. However, for very small pixels (< 0.5 mm) and a required sensor thickness of 3 mm for detection efficiency in CT, there are significant penalties to be paid in the form of energy spectral distortions due to charge sharing between pixels.

Another strategy to shorten the signal duration is the implementation of parallel drift structures [13]. Drift structures allow the collection of electrons from larger volumes on a small anode [14], [15]. With this method a good energy spectral response is preserved and fast signal formation is achieved due to the small anode dimensions. However, drift structures require a larger number of electrodes and smaller anodes than a corresponding electrode configuration using the small pixel effect for the same size detector. More electrodes and smaller dimensions may increase the difficulties involved in electrically connecting the sensor to ASICs.

An alternative detector configuration approach involves multiple stacked layers of segmented detectors with the layers having varying thicknesses. The thickness (X-ray absorption) of the individual layers is optimized for count rate performance and the avoidance of saturation [11]. This approach has a number of disadvantages related to the interconnections between the various layers and readout electronics. In order to keep detector layers close together, connections to the readout electronics need to be elongated causing additional stray capacitances and vulnerability to cross talk. The increased stray capacitance is particularly harmful at high count rates, contributing to higher electronic noise and/or higher power requirements for the readout electronics. Also, detectors with multiple layers may be more expensive to produce.

In parallel with strategies related to detector (sensor) development, there are a number of approaches being investigated to develop new ASIC readout electronics including: a) dense parallel channels of amplification and processing electronics that provide a dedicated electronic channel for each detector pixel, b) parallel electronic channels with an analog portion matching the number of detector pixels and a digital portion such as discriminators and counters common for several detector pixels, and c) electronics that allows for a photon-counting mode at lower rates and a photon integration mode at high rates thus preventing the saturation of the electronics at very high photon flux.

The dense parallel channel method to increase throughput of a detection system is an approach utilizing an increased number of parallel detection and signal processing channels within a given area [10], [16]. However, densely packed multi-channel fast electronics lead to large power consumption and the resulting heat needs to be dissipated to ambient without negatively affecting the detection system. In order to reduce power consumption the shortest possible connections to detector pixels (lowest stray capacitance) are considered. In a case when the detector's segmentation (pixel pitch) is smaller than the intrinsic spatial resolution requirement and is being used merely for increasing count rates per unit area there is the possibility of using common electronics (thus reducing power consumption) such as discriminators and counters that can serve more than one detection pixel without degradation in count rates. Another approach to cope with a very large dynamic range of incoming X-ray photons is the development of electronics with the ability to provide photon-counting information at lower rates and photon integrated information at very high rates, thus preventing saturation of the system [17], [18].

The extremely high X-ray photon count rates impose a number of requirements on the design of the processing electronics and the selection of detector material and type of electrode fabrication. In order to avoid a space charge build up in CZT or CdTe crystals that would lead to distortion and eventual collapse of the electric field, care must be taken that the charge generated by the radiation is removed from the device at a sufficiently fast rate [19]. For this reason hole collection (holes being less mobile than electrons) is critically important. In this respect CdTe has some advantages over CZT as typically lifetimes of holes are significantly longer in CdTe than in CZT. Also, the construction of blocking electrodes in these detectors is very important as the high density of charge induced by X-rays in close proximity to electrodes might otherwise cause an uncontrollable injection of current from electrodes into the bulk crystal.

This paper describes prototype photon-counting energy-dispersive detector modules with the small pixel effect that were designed and fabricated for high count-rate operation and to specifically fit a GE LightSpeed VCT scanner. We present the first clinical data taken to demonstrate the utility of photon-counting and energy-dispersive CT. Finally, we show preliminary results from photon-counting modules with a parallel drift structure which increases the module's count-rate capability.

II. Detector Arrays With Small Pixel Effect

Fig. 1 shows photographs of the anode and cathode sides of a CdTe detector. On the anode side, arrays with 16×16 pixels and a pitch of ~ 1 mm that have been designed and fabricated. The edge and corner pixels are slightly smaller to allow for tiling detectors with preservation of the pixel pitch.

Thickness of the material is 3 mm. On the cathode side there is a common electrode for all pixels in the form of continuous metallization. The detectors were fabricated from high resistivity ($\sim 10^9$ ohm-cm) CdTe with Cl compensation. We also fabricated detector arrays using ($\sim 10^{11}$ ohm-cm) CZT material. Details regarding the detector fabrication, lateral surface passivation and basic detector characterization and testing is given elsewhere [20], [21]. The detectors were specifically designed to provide the spatial resolution and stopping power for X-rays that are typical of integrating detectors used in current clinical CT systems [22]. The design was optimized to take advantage of the small pixel effect in both the spectral response and fast signal formation aspects. Fig. 2 presents a typical pulse shape from a detector pixel obtained in response to a gamma ray photon event from a Co-57 source. Note that the fast portion of the pulse has about a 22 ns rise time.

Fig. 3 shows the spectral response of one small pixel to gamma ray photons from a Co-57 source taken with a customized commercial discrete electronics amplifier and a multi-channel analyzer.

Despite a very short peaking time of 30 ns, an energy resolution of 4.75% (5.8 keV FWHM @ 122 keV) is obtained. The electronic noise is 4.8 keV (FWHM), as measured with an electronic pulser.

III. ASIC Architecture and Characteristics

Our ASIC contains 128 parallel readout channels, each connected to one CdTe pixel. All the channels are equipped with an amplifier, a shaper, two discriminators, two 5-bit digital-to-analog (DAC) converters, and two counters. Fig. 4 shows the basic architecture of an individual channel.

Each channel uses a transimpedance amplifier with an active feedback [23], [24]. The input stage is based on a classical cascode configuration with a NMOS input transistor and a PMOS current source load (see Fig. 5). The input transistor size is optimized to match 5 pF of capacitance from the detector plus the parasitic part. For minimum noise the nominal bias of the input transistor is 500 μ A. The amplifier is AC coupled to a shaper which uses a voltage amplifier in a single ended configuration of two cascaded common source stages. It provides additional amplification and introduces the dominant integration time constant. The gain at the shaper output is about 120 mV/fC and the signal gives a peaking time response on a delta-pulse at the input of about 20 ns.

The shaper is followed by two discriminators with identical circuits. Fig. 6 shows the principal schematic of one discriminator circuit [23], [24]. The threshold is applied differentially producing a DC shift at the output of the differential pair. The first threshold line (VT1) is connected to a 5-bit DAC, while the other (VT2) is common to all the channels. The DAC allows for fine tuning of the thresholds per channel in order to compensate for channel-to-channel variations caused by processing parameter fluctuations. The DACs are linked in a series daisy-chain and the values are programmed by a series shift register. The differential signals are then buffered by two NMOS source followers to the comparator stage that consists of a classical two stage amplifier with very high DC gain. Each discriminator circuit is followed

by an 18-bit counter designed in the form of a asynchronous binary type using static master-slave D-flip-flops in each cell.

The counters are gated in parallel by an external signal and the readout is controlled by a sequential bit register, retrieving the contents of each 18-bit counter fully in parallel. The readout sequencer can operate at 50 MHz, allowing for a complete readout of all the 2×128 counters in about 5 μ s. The full chip architecture is illustrated in Fig. 7.

The peaking time was measured as a function of the detector capacitance [25]. For detector capacitances up to 10 pF, and in response to a near-delta pulse at the input, the signal showed a peaking time of ~ 20 ns. Fig. 8 shows the ENC measured with the analog channel for different detector capacitances [25]. At 0 pF the noise is 430 e- and the noise slope is measured to be 56.1 e-/pF. For the nominal detector capacitance of 5 pF the noise was 711 e-. The measured value is higher than the calculated one, which can be explained by parasitic capacitances that were not taken into account.

The gain was measured by issuing a series of calibration pulses and scanning the threshold. All 128 channels were tested simultaneously at nominal bias currents. Fig. 9 shows the gain distribution across the channels. The average measured gain is 143.5 mV/fC with a channel-to-channel gain variation of the order of 1.5% rms. The comparator offset was measured to 5.2 mV r.m.s., which corresponds to 3.6% of the response equivalent to 1 fC input charge. When the chip was fully operating, a power consumption of 2.1 mW/channel was measured.

IV. Detector Modules for CT

In order to demonstrate the usefulness of photon-counting CT, several prototype detector modules were designed and fabricated with physical dimensions to specifically fit a GE Light-Speed VCT scanner. The modules used for all CT imaging experiments utilize the CdTe detector arrays which make use of the small pixel effect and are able to replace the conventional integrating detector units currently provided with the scanner. Each module included 2 CdTe detector arrays described in Section II and 4 ASICs (Section III) to provide readout from 512 detector pixels. The modules were tested for electronic noise performance and count-rate capabilities. The low discriminator level (VT1) could safely be set above the noise level at 25 keV. The throughput characteristics were taken with a GE TTH Hawkeye X-ray generator as a function of the X-ray tube anode current. The X-ray generator was biased to 140 kVp and X-rays were filtered through a 1 mm thick Cu sheet. Fig. 10 shows the output count rate (taken from the ASIC counter) of a typical detector pixel and its processing electronic channel with the low discriminator level threshold set at 25 keV. The output count rate is fairly linear up to 3.5×10^6 cps and saturates at about 6×10^6 cps. A linear fit to data taken at low count rates shows that the saturation in the output count rate corresponds to about 11×10^6 cps at the input. By using corrections the output data can be made linear well above 3.5×10^6 cps and by using this procedure the useful dynamic range of the detector can be significantly expanded.

V. Prototype Clinical CT System and First Patient Images

The conventional integrating detector modules of a clinical multi-slice scanner were replaced with the photon-counting modules described in Section IV. The detector system counts the X-ray photons transmitted through the patient and sorts the counts into two energy bins. The two bins of count data are processed to produce two sets of raw data (high and low energy), that are input to the CT reconstruction algorithm. The raw data sets are processed with a dual energy calibration technique that enables projection space reconstruction and eliminates the beam hardening effect. The image reconstruction generates mono-energetic equivalent CT images (with a user-selected energy) and “virtual-un-enhanced” images where the iodine-based

contrast media is identified and removed in the reconstruction process (see Fig. 11 and Fig. 12). The images are presented in Hounsfield Units (HU).

The first clinical trial with the prototype system is a prospective study on patients with known carotid artery disease. The study was performed at the Rabin Medical Center in Israel and was reviewed by the hospital IRB committee. The protocol is that of a CT Angiography scan of the carotid bifurcation region using a very low X-ray tube output (approx 1/10 the mA setting as compared to a conventional CTA exam). The axial images are processed to create MIP, 3D, Volume rendering, and curve oblique images. Fig. 13 illustrates examples of the processed images. The clinical images demonstrate the vascular and stenotic elements with good image quality, which is particularly notable in regard to the low X-ray tube current technique used for these scans.

VI. Future Work: Detector Arrays With Parallel Drift Structures

X-ray throughput characteristics of photon-counting detectors are a limiting factor in X-ray flux (setting of current at the X-ray tube) that can be used without saturation of the system. This particularly limits the possibility of using such systems for imaging of moving organs (e.g., heart). Clearly, it is very important to develop detector structures with higher throughput characteristics. Our efforts to achieve higher throughput by applying parallel drift structures have shown the feasibility of this approach. The initial results are presented here.

We performed simulations of the electric field within the detector structure as well as simulations of the resulting output signal for different electrode patterns. We have been exploring the effect of changes to the following parameters: anode size, inner diameter of the contact which surrounds the anode, and the pixel size (thickness and width). Also, we varied the bias voltages applied to two electrodes (V_1 = cathode potential and V_2 = steering electrode potential). The anode potential is by definition kept at 0 V. It was assumed that the maximum voltage that we can apply to the detector cathode should not exceed 1000 V.

Based on these simulations, we chose to construct and test a detector array composed of 256 pixels (16×16) with an inter-pixel pitch of 1 mm and a detector thickness of 3 mm, where the anodes are formed as small 0.3 mm diameter discs surrounded by a common steering electrode and the cathode side is formed as an interconnected grid pattern [14]. Fig. 14 shows photographs of the anode and cathode sides of a CZT detector array with this parallel drift structure.

Fig. 15 presents a typical pulse shape from a single pixel of the drift detector array obtained in response to a gamma ray photon event from a Co-57 source. The steering electrode was biased to 200 V whereas the cathode was kept at 1000 V. The pulse shape is shown in two time scales. The top and bottom drawings show the same pulse in 1000 ns and 10 ns scales per division, respectively. Note that the fast portion of the pulse is about 6 ns.

Fig. 16 demonstrates the influence of different bias voltages applied to the steering electrode on the spectral properties of the detector. Spectra of a Co-57 radiation source were taken from a single pixel of the drift detector array at bias voltages of 0 V, 100 V and 200 V applied to the steering electrode. A bias voltage of 1000 V was kept at the cathode in all the measurements. The main amplifier used a 100 ns peaking time. There is a remarkable improvement in the spectral properties of the detector with the application of bias voltages applied to the steering electrode. This effect is due to the collection of electrons from a larger volume and with much less charge sharing.

Despite the fact that all the spectra shown in Fig. 16 were collected using the same acquisition time, the total number of counts rapidly increased with an increase in the bias voltage applied to the steering electrode. Also, with increasing bias voltage, the majority of counts in the low

energy tail were moved into the photopeak. An energy resolution of 5 keV (FWHM) at 122 keV was measured for the spectrum taken with 200 V applied to the steering electrode (bottom spectrum in Fig. 16). The top spectrum in Fig. 16 models the response of a simple pixel detector with an inter-pixel pitch of about 0.3 mm and detector thickness of 3 mm. In this case of a simple non-drift structure detector, the small pixels allow for very fast signal formation and the possibility of applying more parallel processing channels per detector unit area, however there is a significant distortion in the spectral response.

Fig. 17 presents a pulse from the setup described for Fig. 15 that was amplified using an Ortec Model 579 Fast Filter Amplifier with an external delay line to shape the output pulse. The length of delay line (coaxial cable) was selected to produce a pulse with a peaking time close to 10 ns.

Pulses from the amplifier were fed into a custom fast multi-channel analyzer (MCA.) The resulting spectrum from a Co-57 radiation source is shown in Fig. 18. An energy resolution of 9.8 keV (FWHM) at 122 keV and an electronic noise of 9.5 keV (FWHM) were obtained. An energy resolution of 9.8 keV is sufficient to produce multiple energy images in CT applications. For a 10 ns peaking time (20 ns dead time) for pulses, a paralyzable pulse pileup model allows the calculation of the total throughput (output count rate) of photons [26]. This photon-counting system will be capable of about 20×10^6 counts per second at the output. When the counts are linearized this becomes about 50×10^6 counts per second.

There are several other research groups developing fast photon-counting X-ray imaging detectors [27]–[29], however to our knowledge none have to date achieved maximum output count rates as large as presented here.

VII. Summary

The paper presents novel detector array structures utilizing either a small pixel effect or a drift phenomenon that have been specifically developed for CT applications. These detectors provide a signal response much faster than the transit time of carriers over the whole detector thickness. At the same time the individual detector pixels exhibit good spectral performance at very short peaking times.

We have developed a fast and low-noise ASIC electronics readout that is optimized for amplification and processing of signals from CZT and CdTe detector arrays. Detector modules containing crystals and ASIC electronics were specifically designed to fit into a commercial scanner and replace conventional integrating scintillator detectors with fast and highly efficient photon-counting and energy-dispersive detectors. The output count rate of 6×10^6 cps corresponding to 11×10^6 cps at the input per 1 mm^2 pixel in a fully functional CT detector module has been demonstrated. The first patient CT images were taken with these photon-counting, energy-dispersive detectors. Despite a relatively low X-ray flux, the images demonstrate good image quality, showing great promise for possibly decreasing the radiation dose to patients while also providing tissue decomposition capabilities. Detector arrays with parallel drift detector structures, when optimized and combined with a fast ASIC electronics, can provide a good developmental path for photon counting of the very high X-ray flux ($\sim 100 \text{ Mphotons/mm}^2/\text{s}$) that is currently used in typical CT exams with conventional integrating detectors.

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Fig. 1.

A photograph a CdTe crystal with a 16×16 pixel anode structure at a pixel pitch of ~1 mm on one side (shown on left) and a continuous cathode on the other side (shown on right).

Fig. 2.
A pulse shape measured for a detector pixel.

Fig. 3. Spectrum taken with a single detector pixel in response to Co-57 source using a short shaping time. Electronic noise is represented by pulses injected from an external generator (pulser peak).

Fig. 4.
Channel architecture.

Fig. 5.
Principle schema of the amplifier.

Fig. 6.
Principle schema of discriminator.

Fig. 7.
The full ASIC architecture.

Fig. 8.
Measured ENC [e- rms.] versus input capacitance.

Fig. 9.
Gain versus channel-number.

Fig. 10.

Output count rate taken from a single detector pixel versus X-ray tube anode current. A straight line represents a linear fit to data taken at low count rates.

Fig. 11.

Mono-energetic images reconstructed at 60, 75, and 100 keV (from left to right), respectively. The arrows show the left internal carotid artery. Note: the HU of the iodinated left internal carotid artery decreases with increasing energy.

Fig. 12. Axial images at identical neck level. The left-side image shows contrast-enhanced neck arteries. The right-side image is the result of the un-enhanced process (iodine is identified and removed).

Fig. 13.
(A) 3-D, (B) curved oblique, (C) coronal MIP, (D) volume rendering.

Fig. 14.
Photographs of anode and cathode sides of CZT detector array with parallel drift structures.

Fig. 15.
A pulse shape from a drift detector pixel.

Fig. 16. Co-57 spectra taken with bias voltage (V_2) of 0 V (top spectrum), 100 V (middle spectrum) and 200 V (bottom spectrum) applied to the steering electrode.

Fig. 17.
A pulse from a drift detector array pixel processed by amplifier using a delay line shaper.

Fig. 18.
Co-57 spectrum taken with a drift detector array pixel.