Nuclear Instruments and Methods in Physics Research A 856 (2017) 36–46

Contents lists available at ScienceDirect

Nuclear Instruments and Methods in Physics Research A

journal homepage: www.elsevier.com/locate/nima

Energy-correction photon counting pixel for photon energy extraction under pulse pile-up

Daehee Lee, Kyungjin Park, Kyung Taek Lim, Gyuseong Cho*

Department of Nuclear & Quantum Engineering, Korea Advanced Institute of Science and Technology, 291 Daehak-ro, Yuseong-gu, Daejeon 305-701, Republic of Korea

ARTICLE INFO

Keywords:
CdTe
Energy-correction pixel, High X-ray flux
Photon counting pixel
Pulse pile-up
X-ray

ABSTRACT

A photon counting detector (PCD) has been proposed as an alternative solution to an energy-integrating detector (EID) in medical imaging field due to its high resolution, high efficiency, and low noise. The PCD has expanded to variety of fields such as spectral CT, k-edge imaging, and material decomposition owing to its capability to count and measure the number and the energy of an incident photon, respectively. Nonetheless, pulse pile-up, which is a superimposition of pulses at the output of a charge sensitive amplifier (CSA) in each PC pixel, occurs frequently as the X-ray flux increases due to the finite pulse processing time (PPT) in CSAs. Pulse pile-up induces not only a count loss but also distortion in the measured X-ray spectrum from each PC pixel and thus it is a main constraint on the use of PCDs in high flux X-ray applications. To minimize these effects, an energy-correction PC (ECPC) pixel is proposed to resolve pulse pile-up without cutting off the PPT by adding an energy correction logic (ECL) via a cross detection method (CDM). The ECPC pixel with a size of 200×200 μm^2 was fabricated by using a 6-metal 1-poly 0.18 μm CMOS process with a static power consumption of 7.2 μW/pixel. The maximum count rate of the ECPC pixel was extended by approximately three times higher than that of a conventional PC pixel with a PPT of 500 *n*sec. The X-ray spectrum of 90 kVp, filtered by 3 mm Al filter, was measured as the X-ray current was increased using the CdTe and the ECPC pixel. As a result, the ECPC pixel dramatically reduced the energy spectrum distortion at 2 Mphotons/pixel/s when compared to that of the ERCP pixel with the same 500 *n*sec PPT.

1. Introduction

Photon counting detectors (PCDs), coupled with a sensor via a bump bond, have been developed by a number of groups over the past two decades for medical imaging, because they provide low noise, high resolution, and high detection efficiency for applications [1–5]. A typical energy-resolving photon counting (ERPC) pixel, a component of the PCD, can measure not only the number but also the energy of incident photons. Fig. 1 shows the circuit of the ERPC pixel.

When the X-ray is incident on the sensor, electron-hole pairs are generated in an amount proportional to the deposited X-ray energy [6]. Under an applied electric field, these electrons and holes drift toward their opposite electrodes, respectively. When charges drift to one of the electrodes, bump bonded to an input of an application-specific integrated circuit (ASIC) below the sensor. The drifted charges are converted to electric pulse signals at a charge sensitive amplifier (CSA), the first signal-processing stage at the ERPC pixel. Pulse signals at the output of the CSA, V_{CSA}, are proportional to deposited X-ray energies on the sensor and the pulse signals are compared with different threshold voltages (V_{THL}, V_{THM}, and V_{THH}) at each comparator. Threshold voltages at comparators are pre-defined energy levels used to measure the X-ray energies. N-bit counters, followed by each comparator, increase by 1 up to 2^N-1, when pulse signals exceed the dedicated threshold voltages, which are a boundary for energy bins. The cumulated number at each counter represents the received photons on the dedicated energy bins and the spectral data with each energy bin contain attenuation information of an object from an X-ray tube to the dedicated pixel [7]. Information at each energy bin is used to synthesize a final image or to separate an image of specific-energy range [8,9]. Furthermore, a weighting method for each energy bin was reported to increase the contrast-to-noise ratio (CNR) by up to 90% when the image is synthesized [10–12]. Patient exposure can be reduced through the weighting method with correct energy detection. Therefore, the exact energy detection of photons is critical for an accurate diagnosis at medical applications.

A number of studies have been performed to replace the energy-integrating detector (EID) with the PCD at high X-ray flux, because spectral data cannot be obtained with the EID. However, the long pulse

* Corresponding author.
E-mail address: gscho@kaist.ac.kr (G. Cho).

http://dx.doi.org/10.1016/j.nima.2017.02.084
Received 30 December 2016; Received in revised form 20 February 2017; Accepted 27 February 2017
Available online 28 February 2017
0168-9002/

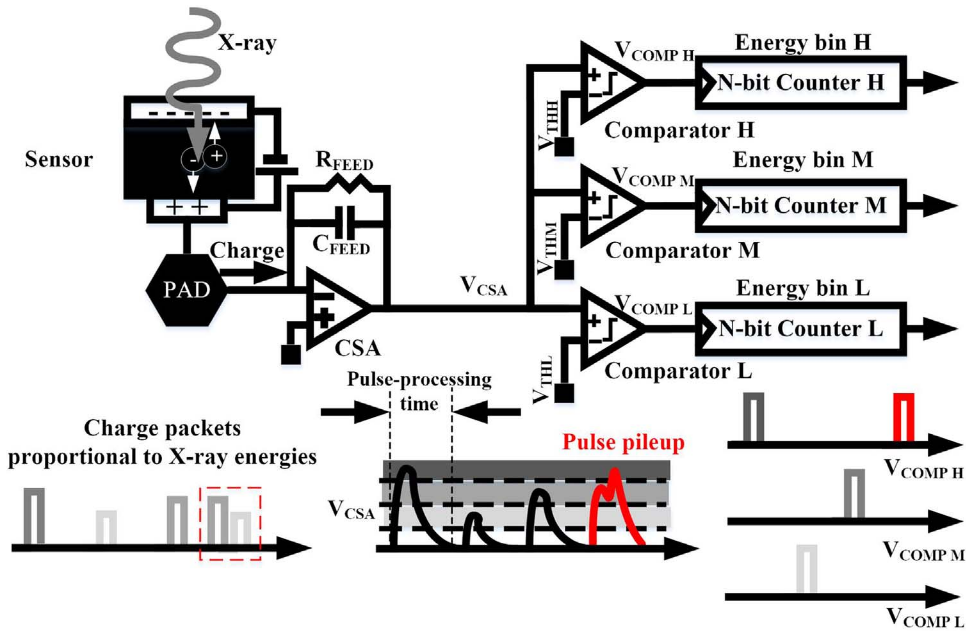

Fig. 1. Circuit topology of an ERPC pixel with three energy bins with pulse pile-up.

processing time (PPT) of the CSA in the PCD causes a train of superimposed pulses, called pulse pile-up, at the high X-ray flux, because the photons are entered before V_{CSA} returns to the base voltage, as shown in Fig. 1. The current PCDs are unable to detect each X-ray energy at pulse pile-up under high X-ray flux due to the finite PPT. Both distortion at the spectrum and the count loss occur due to pulse pile-up. Ultimately, the phenomena deteriorate the SNR of the image and increase the patient dose to maintain the SNR.

Solving pulse pile-up at the ERPC pixel is essential for high X-ray flux applications. 3 methods have been reported to address pulse pile-up before further data processing for correction of pulse pile-up via software.

The first approach to avoid pulse pile-up is reducing the PC pixel dimensions for the pixel to be handled at lower X-ray fluxes in parallel. However, charge sharing, which induces a low-energy tail at the energy spectrum, and characteristic X-ray, which result from the k-escape peak, occur frequently, and thus degrade the energy spectrum again as the pixel size is reduced to under 200 µm [13]. High X-ray flux degrades the energy spectrum whenever a large pixel suffers from pulse pile-up, or a small pixel from charge sharing.

The second method is to attenuate the high X-ray flux with a bow-tie filter before a patient. The bow-tie filter is shaped so as to attenuate most photons such that they do not pass at the periphery, but rather at the center. The X-ray flux right above the PC pixel is reduced after the bow-tie filter so that pulse pile-up can be avoided. However, significant beam hardening at the spectrum occurs at the periphery after the bow-tie filter because X-ray, which have lower energy, are easily absorbed in the filter. The exposure time for the patient must be extended to maintain the SNR due to the reduced photons at the lower energy, and the extended exposure, furthermore, induces motion artifacts, when the region of interest is positioned at the periphery.

The third approach is to shorten the PPT through the reduced value of a feedback resistor, R_{FEED}, in order to avoid pulse pile-up. However, the noise at V_{CSA} is degraded inversely proportional to the feedback resistor value and the loss of charge-to-conversion gain is increased [14,15]. As a result, a sophisticated ASIC design and increased power consumption are required to reduce the noise.

In this paper, we present an energy-correction photon counting (ECPC) pixel with a cross detection method (CDM) to enhance the maximum count rate and an energy correction logic (ECL) for reducing distortion at the spectrum at pulse pile-up instead of shortening the PPT. The ECPC pixel extended the maximum count rate about 3 times maintaining the PPT. And also the ECPC pixel with CdTe (Acrorad) also shows the reduced energy distortion dramatically at 2 Mphotons/pixel/sec X-ray flux compared to the results of the ERCP pixel. The ECPC pixel was fabricated using a 6-metal 1-poly 0.18 µm CMOS process with a pixel size of 200×200 µm^2 with only power consumption of 7.2 µW.

We first describe the proposed ECPC pixel in methods. And the experiment setup to evaluate the proposed ECPC pixel is represented. The maximum count rate and energy spectrum are examined in result and discussion. Finally, our results are summarized with suggestions as a future study in conclusion.

2. Methods

2.1. Proposed ECPC pixel

A ECPC pixel is proposed to solve a count loss and a distortion in the energy spectrum caused by pulse pile-up. The ECPC pixel detects correct photon energies and the number of incident photons without having a count loss even with the presence of pulse pile up. A circuit schematic for the ECPC pixel is shown in Fig. 2.

In the circuit, two logics are added to the ERPC pixel to model the ECPC pixel: the CDM logic and the energy correction logic (ECL). The count loss is alleviated by the CDM logic, while the ECL prevents the energy spectrum from being distorted under the pulse pile-up event.

2.2. Cross detection method logic

When pulse pile-up occurs in a PC pixel with a single threshold, an output of the comparator, V_{COMP}, is triggered only once for two X-ray inputs, as shown in Fig. 3(a). The counter increases the value from a triggering signal and thus a count loss occurs by the pulse pile-up event.

To avoid such loss, a CDM logic was added to the PC circuit. The operation of the CDM logic is illustrated in Fig. 3(b). V_{CCSA} is a copied signal of V_{CSA} with an adjustable delay, T_{Delay}, and an adjustable offset voltage. To add T_{Delay} in V_{CCSA}, an all-pass filter topology was used [16]. The transfer function and phase shift of V_{CCSA} in the all-pass filter were derived by