SPAD based imaging of Cherenkov light in radiation therapy

A Pétusseau[1,2], P Bruza[1], S Tisa[3], S Gioux[2], B W Pogue[1]
[a]University Name, Faculty Group, Department, Street Address, City, Country, Postal Code
[1]Thayer School of Engineering, Dartmouth College, Hanover NH 03755 USA
[2]Telecom Physique Strasbourg, University of Strasbourg, Pôle API, 67400 Illkirch-Graffenstaden, France
[3]Micro Photon Devices srl, Via Stradivari 4, 39100 Bolzano, Italy

ABSTRACT

During radiotherapy, X-ray beams induce Cherenkov light emission in tissue as part of the dose delivery. This light can be used for dosimetry, in order to track and image the dose as it happens. The Cherenkov light levels are in the range of 10^{-6} to 10^{-9} W/cm^2, which makes it challenging to detect in a clinical environment. However, because the radiation is pulsed in 4 microsecond bursts, time-gated acquisition of the signal allows for robust detection, even in the presence of ambient room lighting. Thus, imaging sensors for this application must be highly sensitive and must be able to time gate faster than a microsecond.

In this study, the use of a solid state detector composed of 64x32 single photon avalanche diodes (SPADs) was examined. The advantages of this technology were intra-chip amplification, high dynamic range, superior X-ray noise rejection and fast temporal gating of the acquisition. The results show that the SPAD camera was sensitive enough to detect Cherenkov radiation despite the 4% fill factor. 2D oversampling (x25) was also used to increase final image resolution to 320x160. In this work we demonstrate the SPAD camera performance in imaging Cherenkov emission from a tissue optical phantom and one patient undergoing radiotherapy.

The SPAD camera sensors could be a viable alternative for Cherenkov imaging, as compared to current imaging methods that are mostly focused around image intensifier-based cameras and so have a range of non-linearities and instabilities which could be solved by an all solid state camera sensor.

Key words: Cherenkov imaging, SPAD camera, radiation therapy, dose

1. INTRODUCTION

During radiation therapy, a high energy photon beam is delivered to the patient's body with a linear accelerator (LINAC). Compton electrons generated by the high energy photons, induce Cherenkov radiation in tissues. Cherenkov detection is used for real time visualization of treatment beam on the patient's surface. This Technique permits to improve radiation therapy safety by insuring that the treatment is matching the treatment plan.

Gated acquisition

Since it is required that the patient must be illuminated by room lights in order to allow visual monitoring of the patient, subtraction of the background scene is pivotal for Cherenkov imaging. Radiance of patient skin due to room lights is about 10^{-1} to 10^{-3} W/cm^2. The radiance of Cherenkov emission from tissue irradiated by therapeutic X-ray beams is approximately 10^{-6} to 10^{-9} W/cm^2, depending on the dose rate of irradiation. [1] These large differences in radiance make the detection of Cherenkov emission impossible in the presence of ambient light using standard cameras that integrate the detected light over the whole duration of individual image frame. However, because the radiation from the LINAC is pulsed at very low duty cycle (1:1000 typ.), gated-acquisition of the signal allows suppression of the background signal when the beam is off. Typically, the LINAC radiation consist of microseconds-long pulses, emitted at 360 Hz frequency (pulse-to-pulse period 2.78 ms) Thus, time-gated detection of Cherenkov emission is possible to significantly improve the signal-to-ambient light ratio. As a result, under dimmed room lighting conditions, the ambient and Cherenkov optical signals are nearly equivalent in intensity. [2] The imaging system can either be triggered by the electronic LINAC signals accessed from service panel [2][3], or by detecting stray X-ray photons by a fast detector.

Background subtraction

Using the microsecond camera gating, the background light will appear to be of the same intensity as the Cherenkov light. To extract only the Cherenkov radiation image, the background must be subtracted from the raw Cherenkov image. In order to perform the subtraction, a second frame is captured at a certain delay after the X-ray pulse. Cherenkov and background frame are equalized to eliminate artifacts due to natural room light fluctuation, and subtracted from each other, yielding Cherenkov-only image.

In a standard imaging setup that employs intensified camera, the Cherenkov acquisition must approach video frame rates (5-30 fps). [4] To maximize the efficiency of capturing Cherenkov radiation, the background pulses can be recorded in between two consecutive X-ray pulses. To prevent adding unnecessary noise to the background subtraction, the background pulse duration can be set to be equal or higher than the total duration of X-ray pulses within one frame.

2. MATERIAL AND METHODS

Intensified CMOS (iCMOS) camera are the best option to perform Cherenkov imaging. Nonetheless, intensifier used in iCMOS systems are typically complex systems with a large contribution of intensifier- and CMOS related noise to the photon shot noise of the image. Single-photon avalanche diode (SPAD) detectors could be a good alternative to iCMOS cameras. The goal of this study is to demonstrate the use of a SPAD detector for Cherenkov imaging applications, and to assess its advantages and weaknesses.

SPC3 SPAD camera

SPAD is a solid-state photodetector in which a photoelectron can trigger a large avalanche current [5]. SPADs also have a high time resolution up too tens of picoseconds. [6]

In this project we used a SPAD-based single-photon counting camera (SPC3, Micro Photon Device, Milan, Italy) that is composed of 2048 individual SPADs (64 rows by 32 columns). Each pixel comprises a SPAD detector, an analogue front-end and a digital processing electronics This integrated device provides single-photon sensitivity, high electronic noise immunity, and fast readout speed. The integrated counting electronics allows fast photon counting to 96 kfps, with approximate dead-time of 10ns between frames. [7] Each pixel has three independent counters that enable counting only during specific time-windows for each selected counter. Since this counter can be enabled by and external signal, e.g. an X-ray emission gating signal in our case, the SPC3 can be used for Cherenkov imaging. SPC3 camera is equipped with a USB 3.0 connection to enable real-time data acquisition. To counteract the low fill factor (3%), the chip was equipped with a micro-lens array.

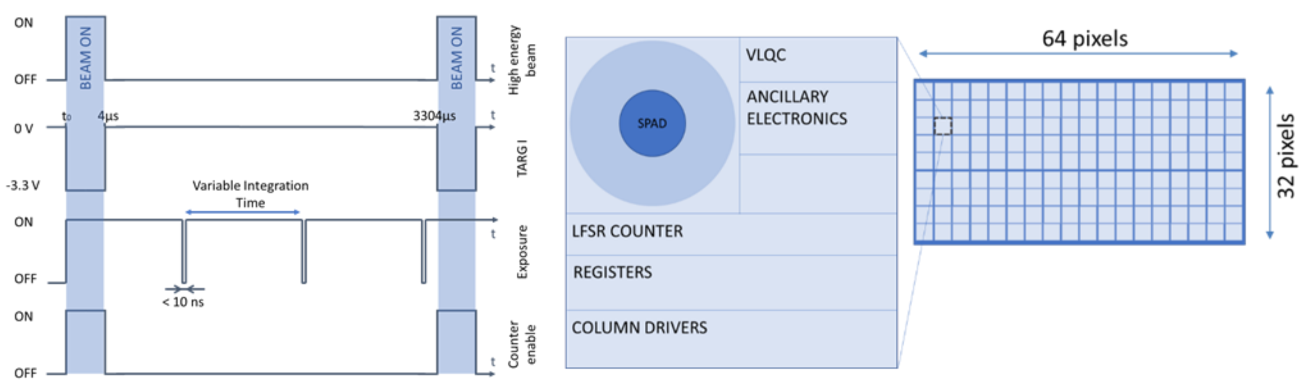

Figure 1: Left: Acquisition timing diagram of the SPC3 camera. Right: SPC3 pixel organisation schematic

Setup timing

In order to acquire Cherenkov light, the SPC3 camera had to be triggered on the LINAC pulses (Clinac 2100CD, Varian Medical Systems, Palo Alto, CA). The trigger signal (TargI) originates from the target producing X-rays from an electron beam. The SPC3 camera needs one trigger signal that starts the acquisition and another trigger signal that is used to gate the individual counters in each pixel. Since both camera trigger inputs require active-high 3.3V pulses

of minimum duration of 40 ns, the TargI signal was digitized by a fast comparator. The comparator circuit converts the TargI signal into TTL (Transistor to Transistor Logic) signal.
The signal delivered by the LINAC is a burst of 3µs pulses at a frequency of 360 Hz. Each counter of the camera measures the number of photons detected during a specified Hardware Integration Time (HIT) corresponding to one frame in our case. The HIT is thus the time during which the impinging photons are counted without resetting the integrated counters. Each frame can only accumulate up to 511 photon detections. Thus, it is possible to accumulate several HIT within one frame (this applies in the case of a high fluence, when a lot of photons need to be counted). In the experiments, the HIT was fixed at the maximum (655µs) in order to reduce the number of blank frames.
Once the acquisition is started by the trigger signal, the chip produced a stream of images in an asynchronous way. The frames were not synchronized with the LINAC trigger signal, but the images were recorded at a frame rate specified by HIT. As the duration of HIT was short relative to the X-ray pulse-to-pulse period, this approach led to a large number of black frames recorded during the image acquisition.

Increase of the camera resolution
In order to increase the resolution of the camera for practical purposes, we used a 2D pixel-shift oversampling technique. As a proof of concept, we mounted both the camera and the lens on a translation stage (LTA HS – Newport, Irvine, USA). A motion controller (ESP300 – Newport, Irvine, USA) was used to automate the process of oversampling. The camera was moving relatively to the lens (Nikon Corp, Tokyo, Japan, f=50mm) with a pitch of 30µm corresponding to the diameter of the SPAD sensitive area. The translation stages have a maximal speed of 5mm/s and an accuracy of ±5.0 µm. The camera was translated across the 5x5 matrix in a shortest path with a delay of 100 ms between each movement. Because of relatively low speed of the translation stages, the total acquisition time is approximately 6 s. While the used translation stages were too slow for practical application, we demonstrate that it is possible to obtain high-resolution images with pixel shift oversampling (figure 2).

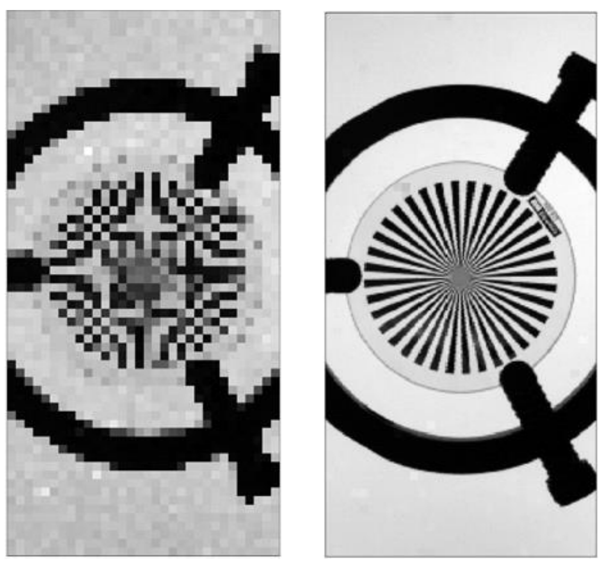

Figure 2: Original low-resolution image versus Oversampled high-resolution image

Phantom imaging
Thanks to the increase in resolution we were able to take images of a water tank phantom to visualize Cherenkov light. The camera was placed on a tripod at 3m in front of the water tank, at the water level to avoid reflection on the water surface. The phantom was irradiated by high energy beams at a dose rate of 600 MU/min. We tested several beam sizes and types (electron and photon beams). In order to image dynamic plans, we acquired a total of 25 videos of the same dynamic treatment, delivered to a white ABS breast phantom. By interleaving the videos together, we could reconstruct a high-resolution video of a dynamic dose delivery.

3. RESULTS

Sensors sensitivity results

Figure 3 shows the sensitivity curve of iCMOS and SPAD camera. On this graph, the number of photons detected by both cameras is plotted as a function of the actual number of photons arriving on the sensors (photocathode in the case of iCMOS). The values from the iCMOS camera were multiplied by an amplification factor corresponding to the LED wavelength (650 nm). Indeed, unlike the SPAD camera that register photons regardless of energy, the iCMOS camera has different response to X-ray photon and Cherenkov photons. The dashed line in figure 3 corresponds to an ideal detector with quantum efficiency of 1 and 100% fill factor. The useful dynamic range of the camera corresponds to the linear range of the sensitivity plots in fig. 3, limited by a noise floor in the low intensity region and by saturation in the high intensity region.

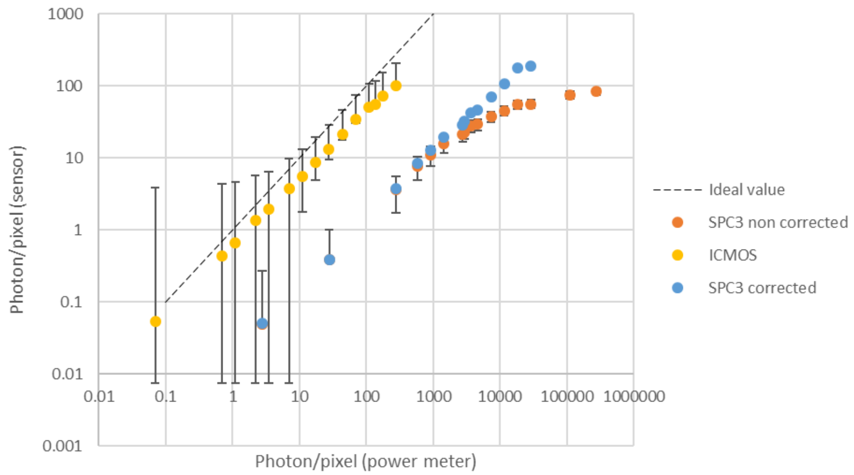

Figure 3: ICMOS and SPAD sensitivity curve with normalized scale. Blue curve corresponds to the dead time corrected signal.

Phantom and patient imaging results

We demonstrated that the SPC3 camera was able to perform Cherenkov imaging synchronized with the TargI signal from the LINAC. Because of its low number of pixels (only 2048 pixels), the SPC3 camera could not acquire images with a good resolution. Thanks to the oversampling method we were able to increase drastically the resolution of the images.

Figure 4 shows the irradiation of a water tank by different beams. Figure 8a, 8b, 8c and 8d respectively show a 10x10 cm^2 photon beam, 2x2 cm2 photon beam, 10x10 cm2 electron beam and 6x6 cm^2 6MeV electron beam. Images are corrected and oversampled. The images show that photons are penetrating much more in water than electrons are.

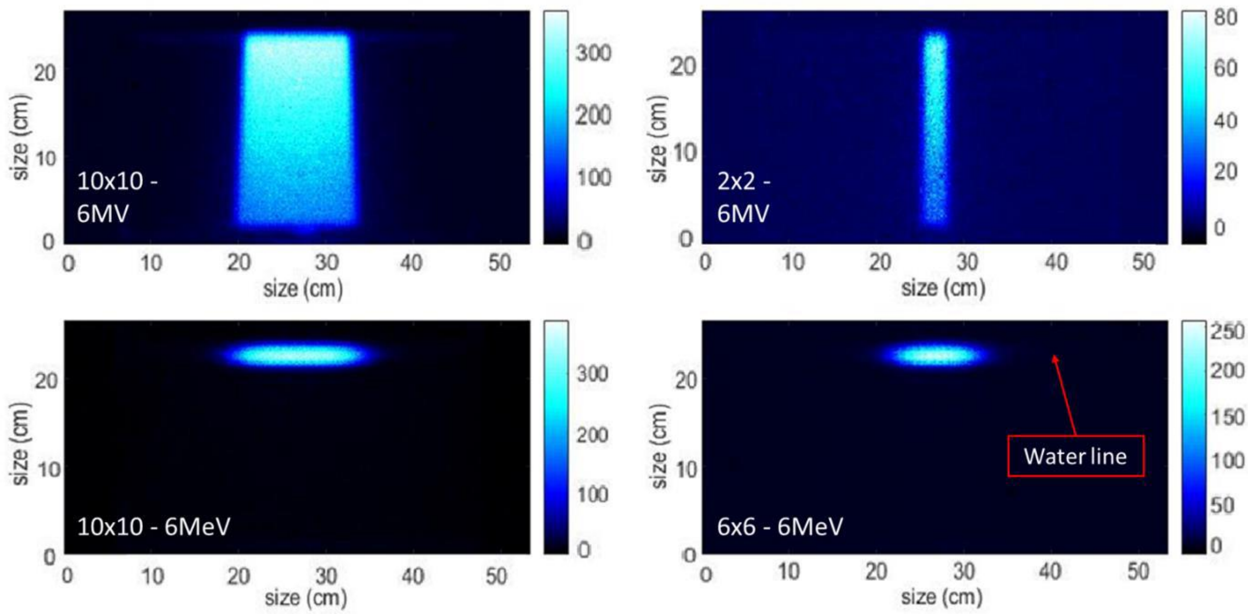

Figure 4: Water tank phantom images with different beam parameters. On top X-ray beams - a)10x10xm b) 2x2cm, on the bottom electron beam - c) 10x10cm d) 6x6cm.

4. CONCLUSION

Cherenkov optical light levels are generally very low, which sets large requirements on the clinical Cherenkov imaging systems, which are designed for routine radiation therapy monitoring. Both single photon sensitivity and time gating are vital for these systems.

Emerging SPAD detectors can provide a very good alternative to cameras equipped with image intensifiers. Their timing characteristics meet the expectations in clinical Cherenkov imaging, end they exhibit excellent X-ray noise rejection thanks to a single bit detection. On the other hand, the SPAD camera has lower photon detection efficiency due to small fill factor, and the pixel count is low. An oversampling setup allowed to increase the resolution of the SPAD camera.

The recent progress in increasing fill factor and pixel count of 2D SPAD detectors promise a more robust, X-ray noise free and compact alternative to the intensified CMOS imaging systems.

5. REFERENCES

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