PIXSCAN: Pixel Detector CT-Scanner for Small Animal Imaging

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Abstract: The aims of the pixel detector CT-scanner for small animal imaging PIXSCAN are to provide high count rate and fast image readout, and to improve soft tissue contrast by energy discrimination, high dynamic range and low noise. As a proof of principle, the photon counting pixel detector built for X-ray synchrotron radiations \cite{1} was used to build a prototype demonstrator of the PIXSCAN device. This demonstrator was used to obtain first images of mice reconstructed by the FDK algorithm \cite{2}. A new photon counting chip with 130 x 130 $\mu$m\textsuperscript{2} pixels (XPAD3) in sub-micron technology has been designed, results from simulations will be presented. We intend to combine XPAD3 detectors with Positron Emission Tomography (PET) to study simultaneous PET/CT co-registration. A dedicated reconstruction algorithm is being developed for a special geometry combining a PET detector ring with a non-transaxial cone-beam X-ray CT design.

I. INTRODUCTION

Photon counting hybrid pixel detectors are being investigated to improve performance over CCD’s and CMOS pixels with regards to image readout duration, noise suppression, dynamic range and efficiency. Figure 1 shows the principle of these detectors. Improved performance is due to the electronic chip connected to the sensor, which includes a full electronic chain for each pixel. Consequently, every pixel can be optimized individually and can count the photons independently. Furthermore we have the choice of the sensor depending on the foreseen photon energy and application. For example, Si sensors are cheap and easy to use, but loose efficiency above 20 keV, even with 1 mm thick material, whereas 0.5 mm thick CdTe is 100% efficient up to 40 keV, thus allowing significant dose reduction. In our case, we chose Si sensors for the prototype demonstrator, but we plan to have CdTe sensors in the final CT-scanner.

The full custom electronics of these new detectors provide an adjustable energy threshold for noise suppression. It can also reject the tail of the energy spectrum that is left after absorption filtering. Furthermore, we can select an energy window by using a second energy threshold. Thus, by making two scans at different energies, we can enhance either bone structures or soft tissues. We propose to take advantage of these properties in order to improve soft tissue contrast of CT-scans. Since 400 to 1000 images are needed for one image reconstruction, the readout speed is of paramount importance. Moreover, the PIXSCAN is expected to provide a non-invasive tool to survey small animals over time. For this,
radiation dose must be kept as low as possible so to avoid drastic damages in the animals. To reduce radiation dose, every photon must be used. This involves ideally to use 100% efficient detectors with no dead time between two consecutive image captures. With hybrid pixel detectors, we can approach this ideal case because we have the choice of the sensor and we can reduce the dead time between two image captures to less than 2 ms.

In medical imaging, one of the main objectives is to provide combined anatomic and functional images. For this, we intend to combine hybrid pixel detectors similar to the ones of the PIXSCAN with the Lausanne ClearPET small animal PET scanner demonstrator developed by the Crystal Clear Collaboration [3]. To obtain true co-registration of PET and CT, both modalities must ideally scan the same field-of-view simultaneously. However, to preserve the sensitivity of the PET scanner, it is better to have a full ring PET detector and not to interleave the PET and CT devices using a partial ring geometry. This would imply to combine a PET detector ring with a non-transaxial cone-beam X-ray CT design for which the standard Feldkamp (FDK) reconstruction algorithm is not adapted. Therefore, a new dedicated cone-beam reconstruction algorithm is under development for this special geometry.

II. THE PROTOTYPE

A. The detector

For the PIXSCAN prototype we use the large silicon hybrid pixel detector (HPS detector), which was already built for crystallography. It has been used on the D2am beam line at the ESRF synchrotron (Grenoble, France) [1]. This detector is made of the XPAD2 electronics chips produced in standard technology (AMS 0.8 µm) [4]. The pixel size is 330 x 330 µm². The XPAD2 includes a 15-bit counter for each pixel and can afford a linear counting rate up to 2x10⁶ ph/s per pixel, corresponding to about 20x10⁹ ph/sec/mm². This chip is low noise (200 electrons RMS on sensor) and allows fine threshold setting. We achieved a threshold dispersion of less than 75 electrons for the full matrix of 24 x 25 pixels for the first engineering run. Although the threshold dispersion was a bit worse for the second production run, we have assembled 8 chips on 500 µm thick Si sensors by solder bump-bonding. Eight of these modules where then mounted in tiles to obtain an HPS detector of 6.8 x 6.5 cm² (Fig. 2).

![Fig. 2. Mounting of the 8 modules to obtain a 44 cm² pixel detector.](image)

B. Data acquisition system

The on-chip readout electronics includes a scan of the overflows from the 15-bit counters of the pixel cells during X-ray exposure. A readout motherboard was built to implement these overflows on another 16-bit counter, one per pixel, in order to achieve a dynamic range of 31 bits (2x10⁹). A fast 33 MHz logic organizes the readout. The readout motherboard includes a memory for 400 frames with 15 bits per pixel (or 200 frames with 31 bits per pixel). The motherboard communicates with a PC using an Ethernet interface.

C. The CT-scanner

The HPS detector is mounted on an aluminum plate, which also supports the readout motherboard on the rear. A small 35 kV X-ray source (Moxtek, USA) with an emission spot of about 800 µm is installed in front of the HPS detector. This is not an ideal choice because of the size of the emission spot and of the limit on the maximum energy of the X-rays. Therefore, it will be soon replaced by a 60 kV X-ray source with a 60 µm emission spot. A remotely controlled plate that can rotate by 0.01 degree steps is placed between the detector and the source. The mouse is installed in a plastic tube positioned at the center of the rotating plate (Fig. 3). Since the mouse is at half distance between the source and the detector, the expected pixel size in the image for this set up is 165 µm (for detector pixels of 330 µm).
III. First Results

A. Determination of the geometry of the detector set up

To reconstruct tomographic images, the acquisition geometry has to be calibrated accurately. For this, we rotate a phantom with several metal spheres. Trajectories of the centre of the spheres are used to determine the exact position of the detectors and of the source with regards to the axis of rotation. An accuracy of half a pixel has been reached by this method. Figure 4 shows a picture of the phantom and a section of the 3D reconstructed balls. The sections of the balls are exact circles as expected when the geometry is correct. Defects result from the gaps between the tiles of the HPS detector. The gaps are horizontal and range from 0.3 mm to 0.9 mm, depending on the position of the tiles with regards to the horizontal plane perpendicular to the rotation axis and containing the X-ray source. We intend to reduce these gaps in the final PIXSCAN design.

B. Images acquisition.

Images of mice have been obtained with the PIXSCAN prototype. For this, mice were installed in a plastic tube in such a way that they could not move.

360 cone-beam projection images (one per angular degree) of the mice were performed [Fig. 5]. Several images without the mouse were also taken to correct for the residual calibration defects.

Cone beam projections were then processed using the RecFDK algorithm developed at CREATIS based on the FDK, to obtain a 3D image of the object. Figures 6 shows a semi-transparent re-projection of the 3D-tomographic images of a mouse (left) and 3D surface rendering for a given threshold (right).
image of the mouse, and a 3D surface rendering for a given threshold. These first results are quite encouraging for the next iteration of the prototype.

C. Effect of the oblique tiling.
By tiling modules, one can build X-rays detectors as large as we want. However, the FDK algorithm assumes that the detector is perfectly planar, so that we could expect some artifacts due to the tiling. To estimate this effect, we have developed a fast analytic simulator of the PIXSCAN geometry.

This simulator was used to produce cone-beam projections of an analytic phantom for a planar detector, and for a detector with oblique tiles and gaps between the tiles. Cone-beam projections were then reconstructed using the RecFDK algorithm and compared by subtraction. Figure 7 shows the comparison of the images obtained with a planar detector and with a detector made with oblique tiles. Differences are quite small and are due predominantly to the small gaps left between the oblique tiles.

D. Effect of the gaps between the tiles
While assembling tiles one cannot avoid dead zones between them. For a 100% efficient X-ray detector, the minimum gap corresponds to the width of the guard rings of the sensor. To estimate the effect of these gaps on the images, we have done a simulation as for the oblique tiling. The result for a large gap (0.5 mm) is shown in Figure 8 (left). In this case, the artifacts are not negligible, but they can be attenuated significantly by interpolating the missing data in the cone-beam projections (Figure 8, right).

IV. PROSPECTS
The next step will be to build a radiation hard 12 x 8 cm² HPS detector with CdTe pixels of 130 x 130 µm². For this, we have designed a new version of the readout chip (XPAD3) in submicron technology (IBM 0.25 µm).

The XPAD3 includes an adjustable double threshold for energy selection and one 12-bits counter for each pixel. As for the prototypes, the readout will run during data acquisition and the time between two images will be less than 2 ms. The design and simulation of the XPAD3 are done and the chip is ready for submission. As an example, the simulated RMS noise is presented in Figure 9.

The CdTe and the bump-bonding are under study in collaboration with the LETI (Grenoble, France). Using 500 µm one can achieves about 100% efficiency at 40 keV (Figure 10).
Fig. 10. X-rays absorption of 500 µm of CdTe compared to the same thickness of Si

The objective is to produce 3 detectors, one for the CT-scanner prototype, and two for crystallography (ESRF-D2am, Grenoble and SOLEIL, Saclay, France). Furthermore, we intend to combine XPAD3 CdTe detectors with the Lausanne ClearPET small animal PET scanner demonstrator to study simultaneous PET/CT co-registration.

E. PET/CT image reconstruction

Using a full ring PET scanner, the X-ray detector and the X-ray source are not rotating in the same plane (Fig. 11), implying a modified FDK formula [5].

Fig. 11. PET/CT geometry.

The results obtained by simulation showed a good image reconstruction in horizontal, (parallel to the source rotation plane) as well as in coronal and sagittal planes when the tilt angle of the source remains limited (inferior to 20°). With larger tilt, if horizontal planes remained well reconstructed, vertical planes suffered from increasing attenuation and geometrical deformations. The circular trajectory is known to be incomplete and therefore leads to approximate reconstruction with symmetric cone beam artifacts when the distance to the source rotation plane is too large. The case of off-centered geometry constitutes even a worse configuration which faces the difficulty of more and asymmetric missing data than standard circular trajectory. Therefore, we are developing new algorithms to reduce these particular cone beam artifacts.

V. CONCLUSION

We have built a large X-ray detector based on hybrid pixel detectors developed for crystallography and small animal imaging. Using this detector, we have built a fast CT-scanner prototype and obtained reconstructed images of mice. To improve the resolution, a new pixel chip (XPAD3) with pixels of 130x130 µm² has been designed. These chips will be mounted on CdTe sensors and associated with a fine spot X-ray source to build a CT-scanner aiming at a better contrast in soft tissues. The same pixel modules will be installed on a ClearPET scanner to build a simultaneous PET/CT machine. This implies a non-standard CT geometry for which new reconstruction algorithms are developed.

VI. REFERENCES