Dynamic X-Ray Imaging System based on an all-solid-state Detector

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Abstract. New digital detector systems based on all-solid-state large area electronics offer a number of advantages for the user, like no image distortions, flat and light weight housing, no veiling glare, and large dynamic range. On the other hand, they require a dedicated image preprocessing to exploit their full image quality. In this paper we address design concepts, performance characterization and resulting image preprocessing aspects of our experimental detector system. The system comprises the detector frontend and a realtime image preprocessing unit with an interface to a commercial digital video system. It is intended for clinical evaluation of this new technology.

Keywords: dynamic x-ray imaging system, all-solid-state x-ray detectors, large area thin film electronics, realtime image processing

1 Introduction

Due to the progress in the field of large area thin film technology a first generation of novel medical x-ray detector systems based on all-solid-state devices is close to market introduction. We have made investigations to a 20 x 20cm^2 detector setup with 1k^2 pixels of 200µm running at up to 30 frames/s [1]. These values come close to the needs of a real product and allow first clinical evaluations in a number of application areas. In this paper we address design, performance and image preprocessing issues of such a novel detector system. The technology chosen offers the possibility of a large detection area in combination with a large dynamic range. Image sizes beyond 40 x 40cm^2 and pixel pitches down to 100µm are within reach. Furthermore the detectors do not show the artifacts found in state-of-the-art systems like geometric distortion, vignetting and veiling glare. In addition, they offer a superior Detective Quantum Efficiency, leaving room for image enhancements. With these properties a single detector system promises the realization of a flat and thin x-ray imaging detector that combines the capabilities for a full size high quality radiography and those for a low dose fluoroscopy system. Systems based on the new detector technology can be considered as a milestone on the track to the digital hospital.

2 Description of the experimental detector system

The experimental system was developed in order to exploit the maximum image quality that could be derived with the thin-film technology. Therefore, we created a very flexible setup based on first principles that can be used for a reproducible performance optimization process. The system comprises the modules shown in Fig. 1:

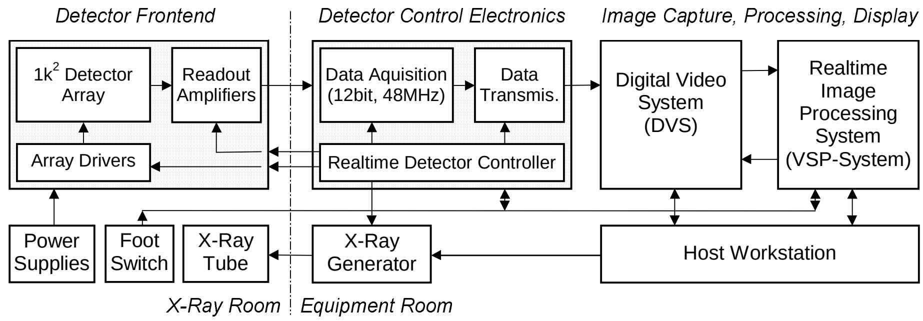

Fig. 1: Experimental x-ray system for first clinical applications

The imaging system is split into two physically separated parts: frontend related components placed in the x-ray room and system control, image acquisition and image processing related parts placed in the equipment room. The whole imaging chain is controlled via a host software running on a Unix workstation. The image sequences are acquired by the DVS and in realtime preprocessed by the Video Signal Processor system (VSP-system). The workstation does image analysis, however, in off-line mode.

The detector is read out by means of charge sensitive amplifiers. The output signals of these amplifiers are digitized by A/D converters with an amplitude resolution of 12bit. Subsequently the image data are multiplexed to a single progressive scan data stream and sent to the DVS at a burst data rate of up to 48Mwords/s.

The DVS allows the recording, archiving and displaying of the generated video sequences. Further it relaxes the input data bursts, transfers the raw images to the VSP system and reads the processed data back. This modality does not replace the entire functionality of a commercial x-ray imaging system, but showed to be sufficient for the clinical investigations.

The VSP system consists of a VME-rack with a 68k control processor board, two VSP boards, two SDRAM memory boards and two I/O boards for interfacing to the DVS. The realtime image processing is carried out by means of the two VSP boards, based on the Video Signal Processor chip (VSP-2 chip). The memory boards, 32Mwords each, are used for realtime storage, e.g. of the correction images. The realtime processor system is capable of processing up to 25 images of 1k^2 pixels per second at a sustained data rate of about 27 MWords/s. The Philips VSP-2 chip is a general-purpose realtime programmable video processor, designed for high performance by using a number of parallel operating pipeline processors and a switch matrix for flexible signal routing. The chip is 12 bit oriented and is running at a clock rate of 54MHz. It has 12 arithmetic logic elements, 4 x 2k dual ported memory and 6 buffer and in- and output elements. The VSP-2 chip contains 1.15 M transistors and is programmable by means of powerful graphical programming tools. Twelve of these VSP chips, configured on the two boards, have been used to realize the corrections. The realtime processing is done frame-synchronously. The total image delay by the preprocessing is less then 0.5 ms.

3 Detector frontend

We applied an approach with an array of photodiodes, Thin-Film-Transistors (TFT) and a scintillator layer on top as the most promising choice. The general architecture of the detector frontend is shown in Fig. 2:

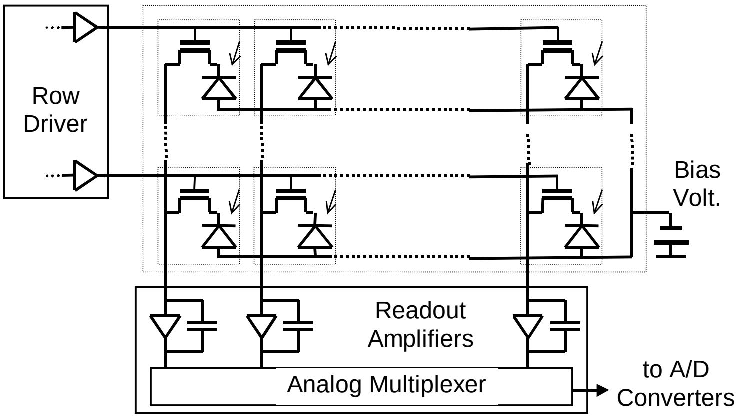

Fig. 2: Schematic of the 2D image sensor

In this architecture all pixels of an entire line are read out simultaneously. The maximum readout speed of the configuration is characterized by the time constant according to $\tau = C_P \times R_{TFTon}$ where C_P denotes the storage capacitance of the pixel photodiode and R_{TFTon} denotes the resistance of the TFT in on-mode.
Our sensor plate is a custom specific development from the Xerox Palo-Alto-Research Center [2]. A dedicated amorphous silicon process for detectors is applied. The top side of the detector plate is covered by an x-ray sensitive CsI:Tl screen which was evaporated directly to the sensor plate. The applied evaporation process guarantees a good trade-off between spatial resolution and x-ray sensitivity.

4 Image quality evaluation

The measured modulation transfer function amounts to MTF(1lp/mm) ≈ 50%. It drops to 18% at 2lp/mm. The Nyqusit frequency is 2.5lp/mm. The detector offers an average electrical signal size of 1750e^-/nGy with respect to incident x-ray dose. The total measured electronic noise amounts to $N_{El} \approx 1200e^-$. Hence, the system yields an S/N-ratio of $S/N_{El} \approx 23.3$ dB at 10nGy dose and the 200µm pixel size.

We calculated the detective quantum efficiency DQE(f) from the MTF(f) and noise measurements. For high doses we get a value DQE(0lp/mm)≈65% at 500nGy. It drops to DQE(1lp/mm)≈52%. We consider these values appreciably high. At lower exposure doses the DQE declines due to the increasing influence of the electronic noise. If we take a value DQE(0lp/mm)=50% as limit, the detector allows quantum limited operation down to a dose of about 4nGy/exposure [3].

5 Image data preprocessing

Different to CCD cameras, the thin-film detectors do not offer image data that can be used without prior conditioning. This preprocessing comprises several basic corrections and associated calibrations of detector imperfections, but no image enhancements. We basically applied offset correction, gain correction and defect interpolation. All fixed pattern noise is removed by this preprocessing. The procedures require dedicated correction sets for each operation mode of the detector system.

As first step a pixelwise offset subtraction is performed. The offset values for each pixel are determined from averaged dark images, so that the subtraction does not introduce additional noise. The offset table has to be updated in regular intervals during operation due to drifts and other thermal fluctuations of the system.

The second calibration is the sensitivity normalization, usually referred to as gain correction. Each pixel is multiplied with its individual gain factor. The factors are determined from the average of equal and homogeneous exposures. This correction also does not introduce additional noise. It is important to notice that the gain showed to be stable over long time periods.

The last procedure is the defect interpolation. Every pixel that showed unusual behavior is marked in a defect map and is replaced in the final images by the interpolation of its neighbors values in either direction. Unusual behavior in first instance means completely insensitive pixels, but also too large or too low offset, more than 25% sensitivity deviation from the average sensitivity and too large noise caused by instability. With these criteria we typically find 0.3% isolated pixel defects and 1-2 broken lines in the current detectors. Only simple interpolation, referencing two suitable neighbors, is applied. Higher order interpolations yielded substitute values for the defect pixels with lower noise. This lower noise can be recognized and is disturbing.

The histogram of a dark image roughly shows a Gaussian-like shape. The values can amount to about 25% of the dynamic range in the highest sensitivity mode. At lower sensitivities the offset distribution becomes relatively smaller. Thus, the quality of the offset calibration is most demanding in the highest sensitivity setting. A detailed analysis shows, that major contributions to the offset values are caused by the preamplifier. The sensitivity variations from pixel to pixel over the total image can amount up to 20%. The named dispersions are not critical, but they limit the available dynamic range in the highest sensitivity mode to about 75% of the possible range. The use of an A/D converter with more than 12 bit resolution would be advantageous.

The term “memory effect” describes a residual image due to charge that is trapped in the photodiodes of the detector after an exposure. The charge release time constants amount up to several seconds. With our array we find about 1.3% residual signal after

1s and still 0.3% after 10s. This means that the memory effect is relevant for mixed mode operation where a low dose fluoroscopy sequence immediately follows after a high dose exposure. According to our current knowledge, the only option for a removal of the after-images is an adequate digital image processing. For this experimental clinical demonstrator a first simple version of memory-effect correction is implemented in order to be able to study the impact of the effects during clinical practice. The correction method is only suited for pulsed fluoroscopy. It is based on dark frames taken in between exposed frames. As a consequence the effective frame speed is halved i.e. from 25 images/s to 12.5 images/s with every other frame exposed. The unexposed images contain the residual signal due to the memory-effect. The correction is done by real-time subtraction of this previous unexposed image from the next exposed image. The method works well, however, it introduces additional noise.
After all corrections have been carried out, the mean pixel value in a region of interest (ROI) is calculated. This mean value is used to do realtime image normalization by scaling the image data to a certain mean level, independent of the applied dose. The normalization is done by multiplying all pixel values with the calculated scaling factor. The mean value in the ROI is also used to generate a control signal to the X-ray generator for dose control during pulsed fluoroscopy mode.

6 Summary

We have described our experimental x-ray system based on a photodiode-TFT matrix detector with $1k^2$ pixels running at 25 images/s. The results show that the enormous potential of the thin-film technology in fact can be exploited in a technical system matched to the medical application. The new detector systems offer the required performance to replace today's film and image-intensifier-camera based systems in all practical cases. In many aspects they do not only offer the replacement, but real system improvements become possible e.g. all kinds of fixed pattern noise can be removed for the first time. The digital detector is an important base for extended digital image processing functionality. Interactive image enhancements will allow the radiologist to match the image presentation to the individual needs exploiting the full potential given by the detector.

7 Literature

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