Flat Panel Detectors for Cone Beam CT: The Complete Package of Large Area, Resolution, Wide Dynamic Range and High Speed

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Abstract

Over the last decade, amorphous silicon based Flat Panel Detectors (FPD's) have matured to the point that they are now rapidly replacing x-ray film and image intensifiers in traditional medical x-ray applications. The advantages of FPD's include high DQE, wide dynamic range, high resolution, no geometric distortion, and high speed image capture. Somewhat unanticipated has been the adoption of FPD's into compact, low cost systems capable of Cone Beam CT (CBCT). The volumetric applications of FPD's already include dental, oncology, and industrial applications and many new areas, such as mammography, are in development. This paper will review the current state-of-the-art in FPD-based CBCT and discuss progress toward sub-second volumetric scanning.

A Short History of FPD-CT

Similar to the digital revolution in commercial photography, digital x-ray imaging is now on the threshold of replacing older analog and film-based technologies. Most imaging modalities in radiology, such as MR, CT, ultrasound, SPECT & PET, are already digital. However, the most wide spread applications, basic radiography and fluoroscopy, are just now making the transition to solid-state, digital, Flat Panel Detectors (FPDs).

At the system level, FPDs offer many of the same advantages of digital cameras relative to their film-based predecessors. Both digital x-ray and digital photography provide: instant image availability, the ability to share the images over a network and the internet, an easy way to make copies and the possibility of image enhancement.

At the detector level, the advantages of flat panel technology include: a small form factor, no geometric distortion, real-time, digital image acquisition, very wide exposure latitude (dynamic range), high detective quantum efficiency (x-ray utilization) and a linear exposure-to-signal response. All of these advantages were anticipated when flat panel development began nearly two decades ago [1]. Less anticipated was the success of FPDs in volumetric CT. There are a number of IIT-based volumetric CT products on the market, but these systems have limited soft-tissue contrast due to the relatively low dynamic range and geometric distortion of the image intensifier / CCD camera. FPD technology overcomes these limitations and as a result there has been a strong resurgence in volumetric CT applications.

One of the first applications of FPD-CT was in oncology, where for many years an IIT-based CT option was available for the Simulator, a piece of equipment used to plan radiation treatment for cancer. Using an FPD, the CT option was expanded from fan beam to cone beam (volume capture in a single rotation) and it became possible to distinguish soft tissue with contrast resolution approaching that of a CT scanner [2]. Another application currently in clinical evaluation is volumetric CT for surgical C-arms [3]. The volumetric dataset offers the surgeon a 3D view into surgical site, during the procedure, without having to transport the patient out to the radiology department. There are a number of flat panel based vascular systems that currently offer rotational DSA, which is a type of cone beam reconstruction applied to the areas of the image containing the contrast injection [4].

While the above are options on equipment dedicated to other more standard x-ray applications, dedicated FPD-CT systems have been the basis for several successful products and some exciting new areas of research. Flat panel technology has enabled compact, office CT scanners that are used in dental implant planning and Ear Nose and Throat (ENT) imaging [5]. The compact size, high spatial resolution and lower cost of FPDs has created a strong interest in CT for mammography screening [6,7]. Clinical trials are in progress [8].

And finally, there has been growing line of research exploring the potential of flat panel detectors in next generation CT scanners [9]. These systems have demonstrated impressive high contrast resolution, greater than 20lp/cm, well beyond what is available in current CT scanners. In addition, the FPD brings fluoroscopy and high quality radiographs to the CT scanner. This research has shown contrast resolution equal to CT scanners, yet with much larger z-axis coverage and finer resolution. In reference [9], the contrast resolution was achieved with electronics extending the FPD dynamic range to 16.5 bits in full resolution (0.194mm at the detector) and to more than 18 bits with pixel binning (0.384mm at the detector). The 0.384mm mode is roughly double the spatial resolution of the best CT scanners.

However, flat panel technology has a number of limitations relative to traditional CT detector technology; specifically image lag, gain stability and frame rate [10]. The greater image lag and history dependent gain stability create image artifacts and can affect the accuracy of the Hounsfield (CT) units. The relatively low frame rate of FPDs creates a problem with intra-slice motion artifacts, despite the fact that an FPD-CT scanner can acquire an organ volume in a total time equivalent to helical CT scanners. In addition to these physics limitations, FPD electronics are not typically designed for the level of linearity found in current CT detectors.

We are investigating these issues using a novel 64 slice CT detector based on flat panel technology. This detector is designed for sub-second volumetric acquisition. The acquisition system is composed of three, overlapping 3.3 x 30cm modules with removable grid and scintillator. In this paper, the 64 slice detector architecture is described, and measurements of the image lag and gain stability at high frames are presented.

Prototype 64 Slice, FPD-CT Detector

The FPD-CT64 detector is composed of 3 overlapping modules, each containing an amorphous silicon array of 64 x 576 pixels with 0.520mm pixels. The electronics is based on our 4030CB, the details of which are reported elsewhere [2]. Each module contains a custom analog readout board connected to the downstream processing hardware of the 4030CB.

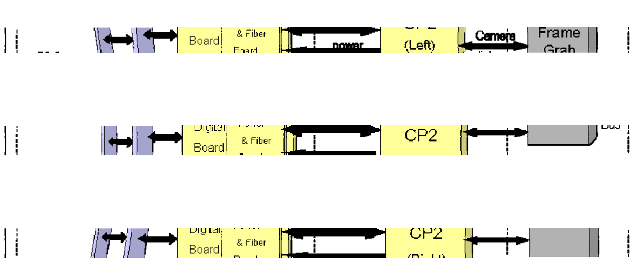
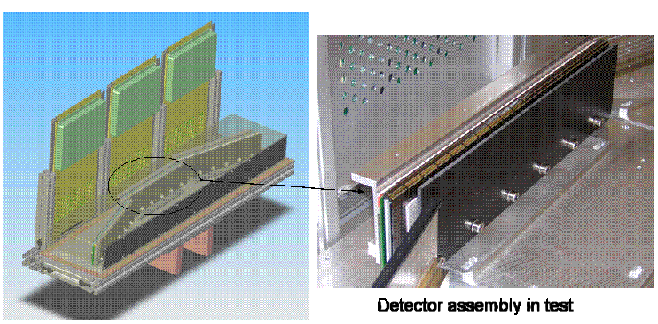


Figure 1. FPD-CT64 architecture

Although the modules are inherently capable of full resolution readout at more than 1700 fps and in 2x1 binned mode at 3000fps, the frame rate here is limited by the 4030CB electronics to approximately 800fps in full field of view, full resolution mode. Currently our array length is limited to 40cm, so three 30cm modules were chosen to create a detector capable of a 50cm scan circle and 70cm aperture. Compared to a tiled structure, the overlapping architecture has the advantages of ease in assembly and no missing pixels. The potential negatives are non-uniform scatter from the adjacent modules and the requirement to remap the data on to an arch or straight line prior to CT reconstruction. Figure 2 shows the detector design concept and the module overlap.



Detector assembly in test

Detector design concept

Figure 2. FPD-CT64 design concept and installed acquisition module.

In addition to the x-ray apparatus for the full 64 slice detector, an optical setup was created to characterize the lag and gain characteristics of the a-Si array, as well as the overall linearity of the detector. An LED array was used to provide a synchronized and well controlled input signal to a single module with the scintillator and cover removed.

Image Lag and the Gain Effect

Using light as an input signal we characterized the response time of the a-Si detector and electronics. The amorphous silicon is known to have high levels of traps which capture charge during the input ramp up and emit charge after the exposure ends. The lost signal at the front-end of an input pulse is referred to as the gain effect, since the apparent gain of the detector increases over the length of the input pulse as the traps fill up and the rate of charge loss decreases. The post exposure ghost signal is referred to as image lag. At lower frame rates the 1st frame lag is typically on the order of 4% and can continue at the 1% level for seconds. The gain effect or trap filling tends to have a shorter time constant. The purpose of our light input experiments was to characterize the gain effect and lag at frame rates relevant to sub-second CT scans, where more than 900 projections are required in less than 0.5 seconds.

The light input signal was monitored using a crystalline silicon photodiode, referred to the norm detector in Figure 3. As shown in Figure 3, the rise and fall of the LED pulse was found to be much faster than the response of the amorphous silicon array. In data shown below, the LED input was asynchronous to the panel readout; however, the similar results were found with a pulsed LED source synchronized to the frame time. Since the electronics response time is on the order of microseconds, the measured detector response is limited by the response of the a-Si pixels.

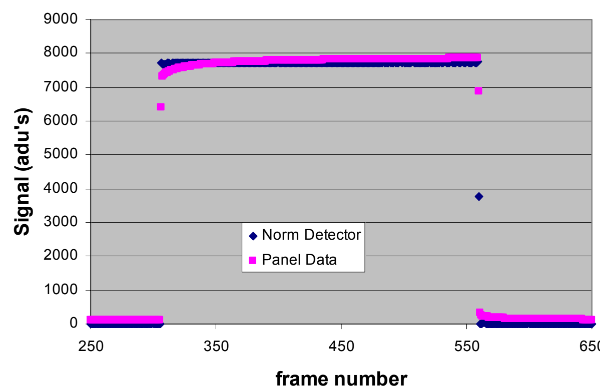


Figure3. Input LED signal and the resulting detector response at 400fps.

Figure 4 shows the post-exposure lag at several frame rates where the signal generated per frame was held constant at 60% of full scale. At each frame rate, the input pulse was 255 frames long. Figure 5 shows the lag dependence on exposure length. For short exposures the lag follows a power law with exponent approximately -0.9. For long exposures there appears to be at least two time constants, one with exponent ~0.35 for the early lag frames and -0.9 for the later lag frames. Figure 4 also shows that the lag is inversely proportional to the frame rate, i.e. proportional to the frame period. Although not shown, the same relationship is seen for short input pulses. In Figure 6 it can be seen that the lag is to 1st order independent of the signal level, which is consistent with results obtained at lower frame rates.

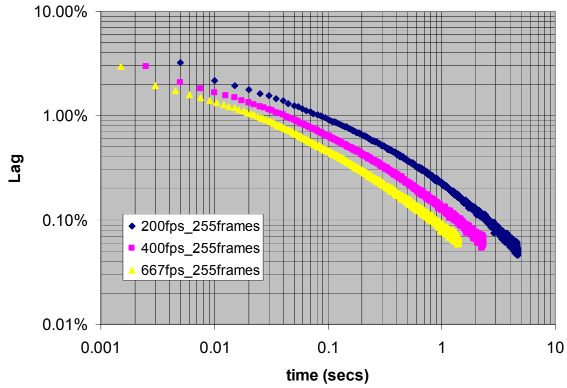


Figure 4. Lag as function of frame rate for a 255 frame long input pulse at 60% of full scale.

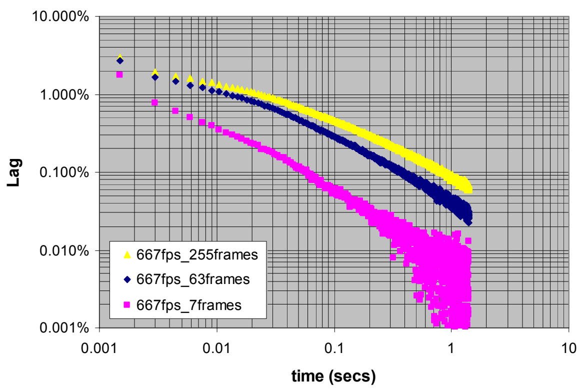


Figure5. Lag as a function of input pulse width. In each case the frame rate is 667fps and the signal level is 60% of full scale.

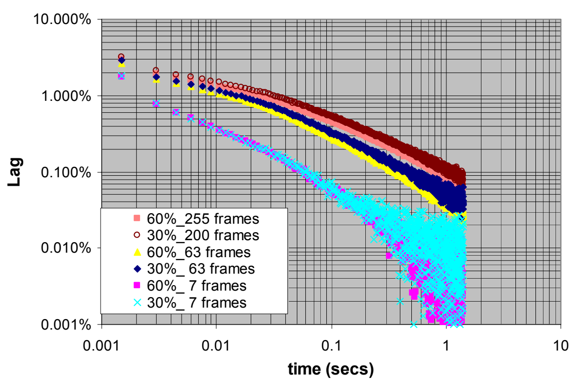


Figure 6. Lag as a function of input pulse width and input signal. In each case the frame rate is 667fps and the signal levels are 60 and 30% of full scale.

The gain effect is characterized with the leading edge of the input signal pulse. Figure 7 shows the change in gain (i.e. signal build up) versus time as a function of frame rate, holding the signal level per frame constant at 60% of full scale. In Figure 8 the same data is replotted versus frame number. In Figure 9 the gain effect is shown as a function the input signal pulse width. The signal per frame is constant across the 3 curves and each is normalized to the maximum signal of the long 255 frame input pulse.

From Figure 7 it is clear that the gain effect can be accelerated by increasing the input signal rate. Since the curves all fall on each other in Figure 8, the total amount of charge required to fill the traps appears to be the same for each fill rate. Figure 9 also suggests that there is a certain amount of charge required to fill the traps

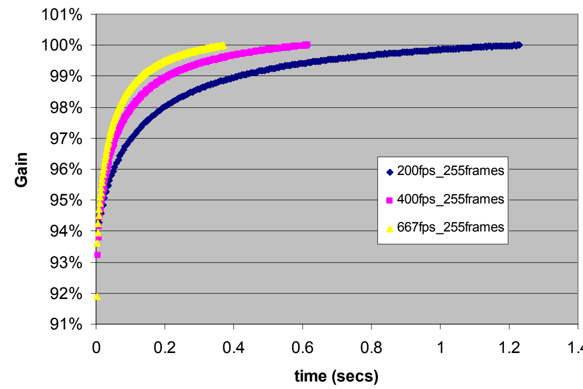


Figure 7. Gain vs. time as a function of frame rate. The signal level is 60% of full scale and the input pulse is 255 frames long for each measurement. Each measurement is normalized to the signal in the 255th frame.

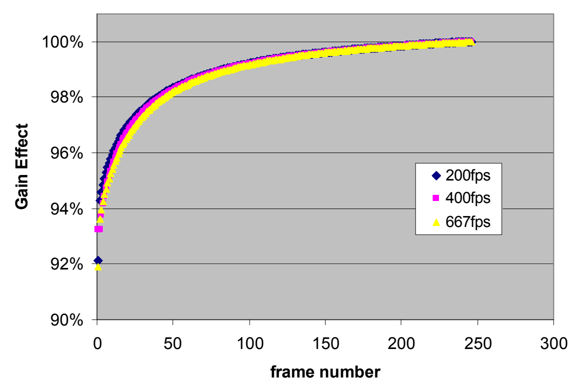


Figure 8. Gain vs. frame number as a function of frame rate for a constant signal per frame 60% of full scale.

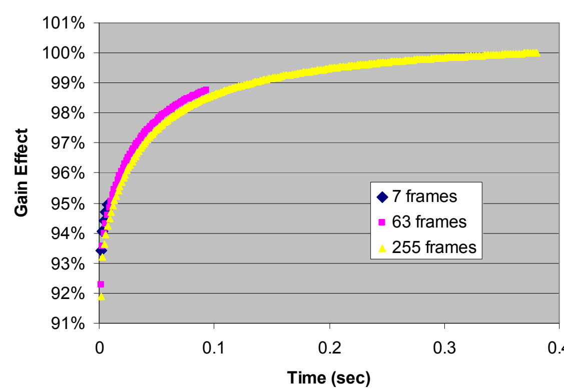


Figure 9. Gain effect versus time as a function of pulse width. The pulse width is varied from 7 to 255 frames and the signal level is set at 60%. All data is normalized to the end of the 255 frame sequence.

Conclusions

As discussed above, the lag is inversely proportional to frame rate, which is consistent with the fact that frame time is the period over which the signal from trap emission is captured. Assuming a constant trap depopulation rate we can estimate the lag at 2000fps, which is the speed at which a sub-second CT detector would need to run. Figure 10 shows the 2000fps lag estimate.

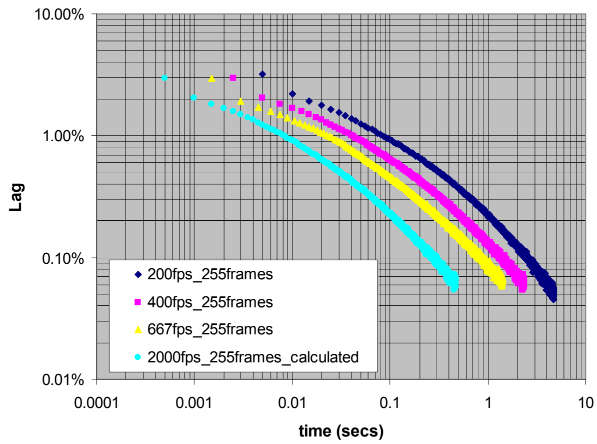


Figure 10. Estimated lag for a long input pulse, 60% of full scale and 2000fps.

In traditional CT detectors with ceramic GOS scintillators the afterglow, which dominates the lag, is on the order of .001% at 300msec post exposure. For the a-Si array running at 2000fps, the estimated lag is ~0.1% at 300msec, or 100 times greater. 300msec is the target time for a revolution in the latest generation of cardiac CT scanners. Here we see that the lag of the array will be present at significant levels over the entire scan. It may be possible to correct for the lag, since it is generally well behaved. To 1st order the lag is independent of signal level. However, as we can see in Figure 6, the lag is exposure history dependent, so that projections from early in the scan will have a different lag evolution compared to later projections.

From the data above it seems likely that the gain effect will also have more influence on the early projections in the CT scan. From Figures 7 and 8 we can surmise that there is a finite amount of charge needed to neutralize the signal loss, which may suggest possible solutions. We are currently investigating ways to mitigate both the lag and the gain effect for CBCT applications.

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Author Biography

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