Digital image capture for high-resolution medical X-ray diagnostics

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ABSTRACT

We have demonstrated a high resolution (10 micron) X-ray scintillator plate as part of an indirect X-ray detection system. Scintillator plates are typically integrated with a 2-dimensional array of photodiodes based upon amorphous Si. This paper describes an alternative digital capture system that leverages low cost CCD/CMOS cameras. Our detector has a broad set of potential applications, however the initial target application is mammography. Full-field mammography mandates an imaging area of 180mm x 240mm or larger. Very large CCD/CMOS sensors have recently been developed for high resolution cameras, such as the 250-pixel Canon camera which has sensor dimensions of 202mm x 205mm, and could conceivably be matched to our high-resolution scintillator plate without any intervening optics for magnification. However, such large format CCD/CMOS sensors have limited availability because of low production yields and high cost considerations. On the other hand, small form (36mm x 24mm) and medium format (44mm x 33mm) CCD/CMOS-based photodiodes have become widely available at low cost due to their applications in the large markets of mobile devices and consumer cameras. We have therefore developed a simple optical scheme for utilizing four small or medium format CCD/CMOS cameras to capture a larger, high-resolution image. Current systems employed in screening mammography resolve tissue features of 75-100 microns. Suspicious features found during preliminary mammographic screenings are further investigated during diagnostic mammographic tests which use a high-resolution detector that is focused over the suspicious lesion. Typically, an area less than 100mm x 80mm, the current maximum size of our high-resolution scintillation plate, is interrogated. We show that diagnostic mammography, over an area of 100mm x 80mm, could be performed using our system with a feature resolution down to 7 microns.

1. INTRODUCTION

About 300 screening mammograms are performed on patients at the Massachusetts General Hospital every day, a typical volume for a tertiary care hospital. Patients are asked to return for a further diagnostic examination if suspicious conditions are detected during such screening mammography, which is performed at a resolution of approximately 75-100 microns over an 18cm x 24cm area. Diagnostic mammography, which is performed after initial screening, is generally done over a smaller area, but at higher resolution. In this paper, we present a system for diagnostic evaluation that could resolve a suspicious lesion down to less than 7 microns over an area 10cm x 8cm.

There are two general categories of X-ray detectors: direct and indirect. In direct detectors, the incoming X-ray photons create an electron-hole pair in the selected absorber material. Absorber materials are typically semiconductors, and can thus be structured (for example as a transistor) in order to detect the electronic carriers which are created. The image is constructed directly from the semiconducting transistors, or pixels. Most direct detectors used for mammography are based upon amorphous semiconducting selenium, and have a resolution of about 75-100 microns.[1]

The larger category of X-ray detectors are indirect detectors, in which the X-ray absorber is a scintillating material, i.e. X-ray absorption results in photoemission. The scintillating plate must be appropriately coupled to some type of photodetector in order to generate an image, as discussed in more detail in this paper. Scintillation-based detectors face two challenges:

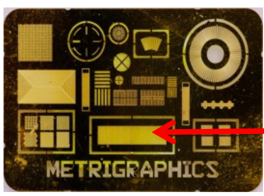
the dose rate or overall efficiency of the detector; and resolution, which is strongly impacted by the 4 pi radial distribution of the photoemission.

Our group of research collaborators have recently developed an X-ray scintillation detector which can achieve high resolution, and is the subject an accompanying paper in this symposium. The present paper addresses the topic of image capture using that detector, using mammography as an example of a diagnostic application in order to focus the discussion.

2. THE SCINTILLATOR

It is well known that “Scintillator-based detector arrays have a limited resolution due to a lateral spread of the scintillating light within the converting layer.”[2] This is why X-rays have such poor resolution. Some of our work has focused on solving this problem.[3,4]

15” x 1.0”



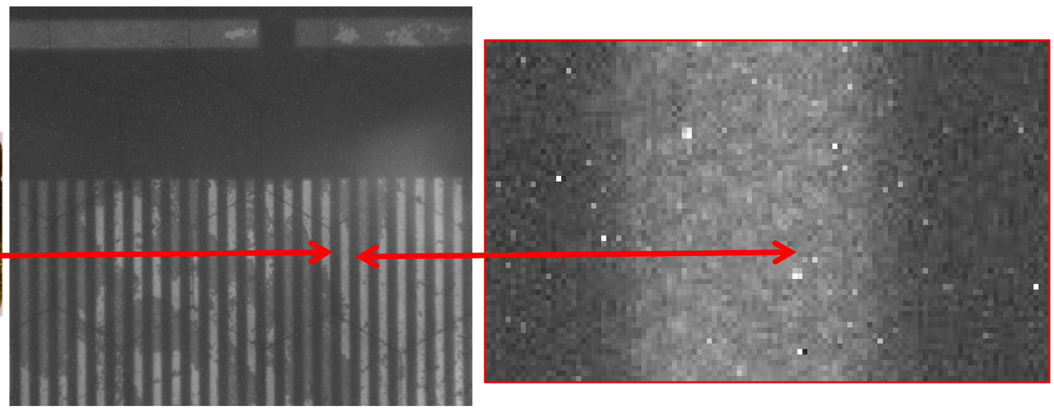


Fig. 2. X-rays: 50keV

Table 1. Incom Coherent Bundle Sizes

Size	μ
8” x 8”	20
8” x 8”	10
5” x 5”	10
53mm x 50mm	7
20mm x 20mm	5

In a previous paper, we have shown that the resolution of X-rays is 10 microns (50 line-pairs/mm)[5] at a visibility of 10% on the MTF diagram. See Fig. 2 and Fig. 3. Incom furnishes the coherent bundles (Table 1), and Lawrence Livermore National Laboratory supplies us with the scintillating bismuth polymer (30,641 Photons/keV).[6,7,8] We infiltrate it uniformly into the small capillaries. The coherent bundle (20mm x 20mm, 5 micron) has 20,000,000 capillaries.

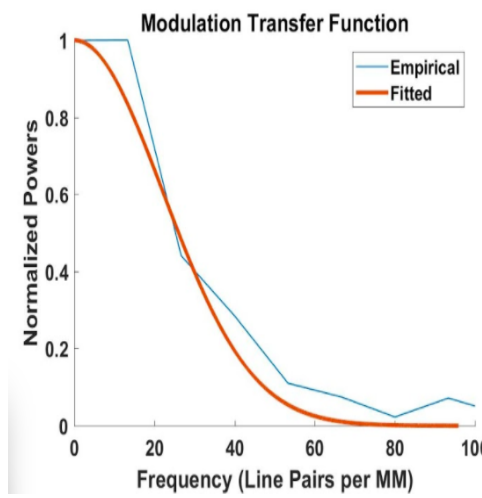


Fig. 3. MTF of Incom 10-micron coherent bundle infiltrated with scintillating bismuth polymer with 50keV X-ray

These scintillator-infiltrated coherent bundles have demonstrated that the resolution for X-ray detection is limited by the capillary pore size. Although we have concentrated on mammography, many other applications are feasible. For example, one can use our detector for imaging atherosclerotic plaque at high resolution, without being affected by calcium blooming that obscures fine features of the plaque due to poor resolution. The inspection of the plaque by X-rays will lessen the chance of coronary infarction.

3. THE PHOTODETECTOR

Most indirect digital X-ray imaging systems integrate a photodetector with the scintillator plate, usually using thin film transistors (TFTs) adapted from large digital display technology. This approach is particularly suitable for current generations scintillator technology, as the available TFT pixel density of around 100 microns matches the resolution capability of current scintillator plates. With the low cost of TFT arrays, this enables the current generation of cost-effective detector systems.

But as we increase the resolution of the scintillating plates, we require much higher pixel density TFT arrays (i.e. with a transistor pitch less than the target resolution), and these are not readily available. It therefore forces us to look towards silicon-based CMOS photodetectors, with their much higher pixel density. CMOS photodetector technology is driven by the demands of smartphone cameras and digital photography. At the very high end, 120 MegaPixel (MP) cameras are just becoming available, approaching the 200+ MP that we would require to achieve a resolution below 10 microns. However, due to the dramatic reduction in process yield with increasing detector area, these photodetectors with pixel counts above 120 MP are prohibitively expensive, and not available in large quantity.

In light of these factors, we have explored the simple question of whether we could use the low cost CMOS detectors of smaller pixel count that have been developed for high volume smartphone and high volume digital cameras. A simple calculation of resolution requirements in relation to the pixel counts of the low cost CMOS detectors shows that we could utilize either "small format" or "medium format" CMOS 24 MP detectors (see Table 2) to achieve 7-micron resolution, but only if we utilized multiple detectors.

Table 2: CMOS photodetectors developed for high volume digital cameras

Sensor	mm	area
H6D-100c (Medium Format)	53 x 40	84.8
H8D-50c (Medium Format)	46 x 24	58.0
35 mm (small Format)	36 x 24	34.56
APS-C	23.4 x 15.8	14.49

Of course the use of multiple detectors in a "tile" arrangement flat against the scintillator would yield two major problems: 1) A large number of CMOS detectors would be required, e.g. at least 18 small format photodetectors for a scintillator size of 10 x 10 cm; and 2) The "tiling" would leave "dead space" between the detectors, which would be unacceptable to radiologists.

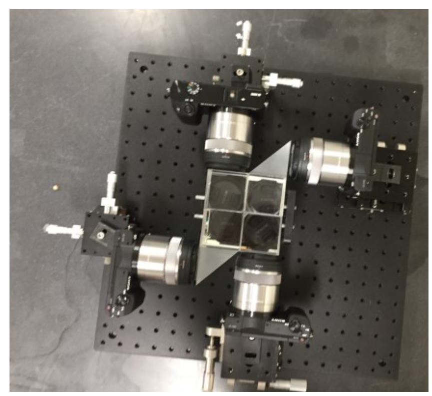


Fig. 4. "Lazy Susan" configuration of mirrors and cameras, 18"

As a result, we have devised a relatively simple strategy in which we utilize an optical camera system. Simple calculations indicate that 4 cameras with small format 24 MP CMOS photodetectors (e.g. the Sony mirrorless camera a5100 APS-C with macro 3.5 lenses) would be required to image the individual channels of a 10 x 10 cm scintillator plate at a resolution of 7 microns (or 70 lp/mm). The need to ensure overlapping images is addressed by using four orthogonal mirrors, and the mirror and camera configuration shown in Fig. 4. Standard image analysis software could be used to "stitch' the four images together into a single image. The system has been assembled, and the imaging system described has been initially tested using a phantom, and complete characterization is underway.

4. APPLICATION TO MAMMOGRAPHY

While the detector system has a variety of potential applications in medical imaging, we discuss the application to mammography in this section. Ductal Carcinoma In Situ (DCIS), a common lesion that is the target of screening mammography, occurs in the milk ducts and cannot be resolved directly by X-rays given their current resolution. Therefore, much of mammography screening is directed towards detecting the collateral effects of DCIS such as micro calcifications and architectural distortion. Thus in our proposed approach, if collateral effects are detected using the standard detector, and DCIS suspected, then a second high resolution screening would be performed using the new scintillator system. Knowing the location of the suspicious region (i.e. from the initial screening) allows the high resolution secondary screen to be performed using a small detector, i.e. the 10 cm x 10 cm size of our prototype device.

While the new scintillation detector has been shown to deliver resolution down to below 10 microns, and a method to capture the photoemission from the detector plate using low cost detectors has been devised, we are well aware that there remain at least two major issues:

1. The higher resolution of the detector will also require modification to the X-ray source to minimize beam spreading
2. Any new detector must not increase the required X-ray dose rate for a given image quality. The dose rate requirements are the focus of the current research.

CONCLUSIONS

In a typical mammography center, about 4-6% of the patients are asked to return for a subsequent cancer screening. The mammograms are performed on a detector system with 70-100-micron resolution (e.g., Hologic at 70 microns, or GE at 100 microns). In conjunction with a new high resolution scintillation detector, we have devised a way in which the photoemission of the detector can be imaged using low cost, Small Format-7.2cm x 4.8cm, or Medium Format-10.6cm x 8cm CMOS photodetectors. Such a setup can provide 7-micron resolution which would be useful in diagnostic mammography for further interrogating a lesion found during a screening mammography.

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