Low cost digital radiographic imaging systems: The x-ray light valve

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ABSTRACT

In recent years, new x-ray radiographic systems based on large area flat panel technology have revolutionized our capability to produce digital x-ray radiographic images. However, these active matrix flat panel imagers (AMFPI) are extraordinarily expensive compared to the systems they are replacing. Thus there is a clear need for a low cost digital imaging system for general applications in radiology. Different approaches have been considered to make lower cost x-ray imaging devices for digital radiography, including: reducing the cost of existing flat panel systems, scanned projection x-ray, an approach based on computed radiography (CR) technology, and optically demagnified x-ray screen/camera systems. All of these approaches are quite expensive and none have the image quality of AMFPIs. We have identified a new approach – the X-ray Light Valve (XLV). It combines three well-established technologies: an a-Se layer to convert x-rays to image charge, a liquid crystal (LC) cell, and a scanned digital readout. This device achieves our goal of immediate readout with image quality comparable to an AMFPI, while keeping costs low. The XLV system has been shown to have all the properties required for general radiography.

Keywords: Digital radiography DR, Direct conversion detectors, Digital x ray imaging radiography DX, Detector technology DET

1. EXISTING APPROACHES FOR LOW COST DIGITAL RADIOGRAPHY

We make a critical evaluation of the different approaches to digital radiography to identify a low cost x-ray imaging device for digital radiography. The criteria are image quality and potential for being made at low cost to permit their use for general applications in hospitals, clinics and health facilities in developing countries.

Different approaches have been considered in the literature to make a low cost x-ray imaging device for digital radiography:

1.1. Reduce the cost of existing flat panel systems [1, 2]

Flat panel technology was originally developed using an indirect conversion approach. A major reason for developing the direct conversion flat panel approach was that the fabrication method for the active matrix arrays can be made compatible with the standard *a*-Si:H TFT manufacturing processes used to make displays. Although this approach permits a lower cost than indirect conversion AMFPI, it cannot be regarded as a low cost system.

1.2. Scanned projection x-ray

One way of producing a large imaging field is to create an image receptor that is essentially one-dimensional and acquire the second dimension of the image by scanning the x-ray beam and detector across the patient [3]. A significant advantage of this approach is the excellent rejection of scattered radiation from the resulting image. However, this method has drawbacks: significantly increased mechanical complexity due to the requirement for an accurate system to synchronously scan the x-ray collimator and detector, and increased heat loading due to inefficient use of the output of the tube. These factors make the system cost similar to that of AMFPIs.

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1.3. Photostimulable phosphor based systems [4]

Probably the most successful detectors for digital radiography to date have been photostimulable phosphors, also known as storage phosphors. These have been used in computed radiography (CR) systems with plates which are carried to a large reader. In order to modify a CR system for immediate readout, the reader must be made very compact and made as a line scanning system. These systems have very poor image quality, requiring approximately four times as much radiation to achieve the same image quality as AMFPIs [4]. The reason for this is the combination of the relatively poor absorption of CR screens having acceptable resolution with the secondary quantum sink in the CR reader. It is possible to make higher image quality integrated CR systems by using structured phosphors, transparent phosphors, double sided readout and/or linear scanning arrays but only at the cost of increased complexity. In summary, the lower cost CR systems do not have good image quality and the higher cost systems can only approach and never equal the image quality of AMFPIs.

1.4. Other approaches using amorphous selenium (a-Se)

There has been a separate effort to investigate readout systems based on photoconductors. A digital method based on xeroradiography was developed which was read out using a scanning laser that stimulated an optical phosphor incorporated into the toner particles [5]. The problem with this approach was that it was impossible to adequately remove the toner to reuse the plate.

The most straightforward approach is to use an array of electrostatic probes held sufficiently close to the a-Se surface (~100 μm) to maintain adequate resolution. The approach has the advantage of very low electronic noise. However, its only practical implementation used a drum coated with a-Se. However, the method is extremely complicated, requires accurate positioning of an array of electronics over a spinning drum, hence it is expensive to make.

Optical discharge methods have the advantage of high resolution that is essentially limited by precision of the focusing of the readout light onto the a-Se surface. There are, however, problems with noise sources not present in the electrometer readout method. The use of segmented readout electrodes or individual electrodes reduced amplifier noise by lowering the capacitance loading the preamplifier. However, no laser discharge method has yet been devised which eliminates the additional noise due to the bias charge that is necessarily read out in conjunction with the signal charge. As a result, these detectors have a poor signal to noise ratio [6].

Thus, all these approaches have significant image quality problems and are quite expensive to make.

1.5. Optically demagnified x-ray screen/camera systems

Optical coupling of a screen to a CCD (or CMOS) digital camera is the most intriguing in its basic simplicity – to combine a screen (a low cost, readily available item) with a scientific CCD camera. Unfortunately, despite many individual attempts to use this approach, it has not been possible to achieve a low cost system. In fact most, if not all, of the commercial systems, as well as being high cost, also have poor DQE [8]. The physical reason is straightforward. It is not possible to make an optically coupled system with a significant demagnification without a serious secondary quantum sink. We can think of an indirect conversion AMFPI as the limit of such a system when the optical detector and the x-ray screen are of the same size, then contact between the elements is possible and by elimination of the lenses, essentially 100% optical coupling is possible and with it, high DQE. However, the low cost version with a simple low cost camera is doomed by the physics to have low image quality.

2. NEW APPROACH – THE X-RAY LIGHT VALVE SYSTEM

All of the approaches above are quite expensive and none have image quality matching that of AMFPIs. Therefore, a revolutionary approach to achieve a low cost, high quality digital x-ray detector system is needed. We have identified such an approach, the X-ray Light Valve (XLV) [9, 10]. It combines three well-established, low cost technologies: the photoconductor a-Se, a liquid crystal (LC) cell, and a digital scanned readout. It functions in a straightforward way: the a-Se and LC cell are combined to make an XLV, in which an optically visible image is generated upon exposure to x-rays; the optical image is read out with a digital scanning device. This achieves our goal of immediate readout with image quality comparable to that of an AMFPI while keeping costs very low. A related approach, using a polymer dispersed liquid crystal (PDLC), has also been considered for digital mammography [11, 12].

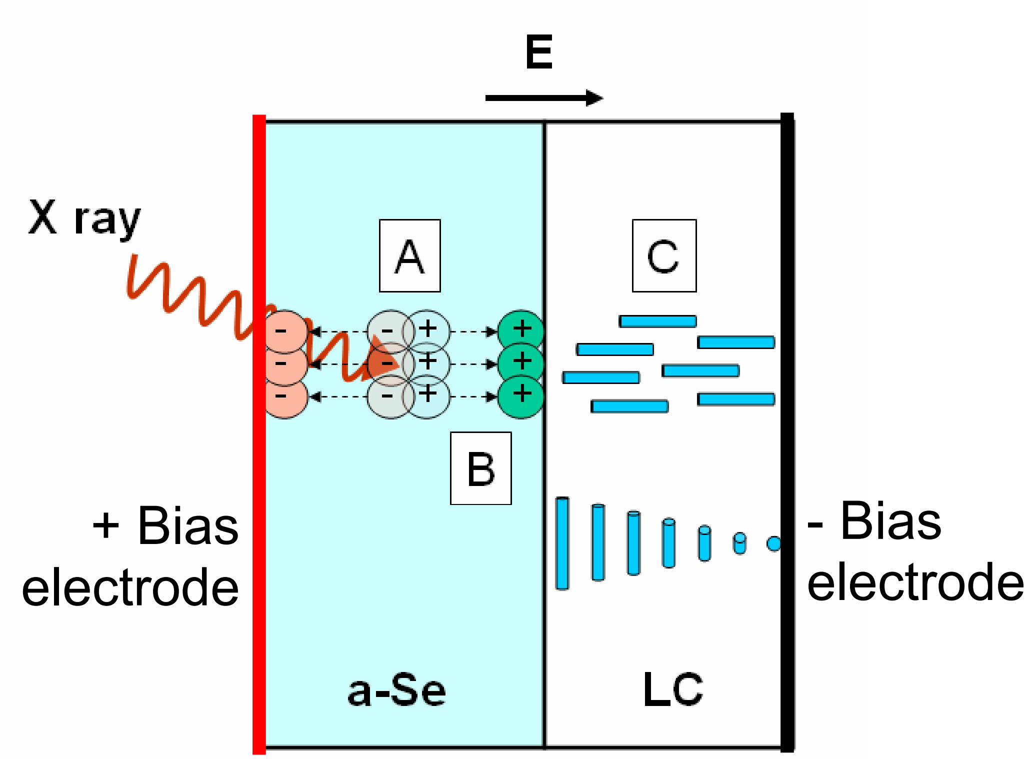

Figure 1: Structure of the XLV. An a-Se layer is sandwiched with an LC cell, and is externally biased using biasing electrodes.

2.1. Operation

The operation of the XLV system can be broken down into three steps: exposure, readout and reset. These are briefly described here to provide context for the more detailed discussion of the system design and properties which follows.

2.1.1. Exposure

When an x-ray is absorbed in the photoconductor, electron-hole pairs (ehp) are released (A in Figure 1). A large applied electric field E guides the electrons and holes to opposites surfaces of the a-Se layer (B in Figure 1). The charges follow the electric field lines, which ensures that there is very little lateral spread as the charges move and thus the charge image collected at the a-Se-LC interface faithfully reproduces the absorbed x-ray intensity pattern.

2.1.2. Readout

The large applied bias electric field applied via the bias electrodes used to guide the ehp is removed. The charges which have been collected and trapped at the interface remain there and create spatial variations in the electric field across the LC cell. An LC cell with no applied field contains LC molecules oriented in a specific direction. The rod-like LC molecules have an anisotropic polarizability, so applying a field to the cell causes the LC molecules to rotate to more closely align themselves with the electric field lines. The larger the electric field in a particular region, the more LC molecules will rotate (C in Figure 1).

The LC molecules are birefringent, and the cell is designed so the polarization state of light passing through the cell is modified by an amount which depends on the orientation of the molecules. An external light source is used for readout, and light is either transmitted through or reflected from the cell. Reflective mode will be described here. A polarizer is placed on the LC side of the structure and the changes in the polarization state of the light are converted to changes in reflectance.

To summarize, variations in potential exist where x-rays were absorbed, causing spatial variations in the intensity of light reflected from the LC cell, giving an optical representation of the x-ray image. This image is scanned in using external illumination and a digital imaging device. It should be noted that the wavelength of the readout light is chosen so that it will not erase the image at the a-Se-LC interface.

2.1.3. Reset

A key feature of the XLV is that the charge image is stable at the a-Se–LC interface, allowing the readout to continue over minutes if desired. As a result of this long lifetime, it is necessary to eliminate the remaining charge at the interface before a fresh exposure is made. To do this, the photoconductor is flooded with actinic light. The charges produced will move towards the interface to neutralize the remaining image charge. The XLV is then ready to acquire a new image.

2.2. Properties of the XLV system

We have elsewhere theoretically analyzed the properties of the XLV [10] and have shown the XLV to be able to meet clinical radiography requirements: x-ray quantum noise limited operation, high x-ray sensitivity, and adequate contrast over the exposure range of interest. It has high resolution because of a combination of properties: the XLV device is not pixelated, the a-Se used is an electrostatic detector in which little blur occurs [13], LC cells inherently have resolution on the order of a few microns, and available digital optical imaging devices have ample resolution. Therefore, high image resolution can be achieved with photodetector layers thick enough to provide high quantum efficiency. The XLV exhibits low noise and x-ray quantum noise limited operation because it requires no external electronics. The fundamental system source noises and the quantum efficiency of the proposed system are modeled elsewhere [9, 10]. By proper choice of the XLV design and operating parameters, x-ray exposures within the clinical exposure range can be optimized for desired imaging tasks [10-12].

In practice, the LC cell image quality and contrast ratio is also affected by: the quality of the polarizers and the LC alignment layers, readout light intensity, uniformity and spectral distribution, undesired light polarization effects (*e.g.* due to scattering at interfaces and in the bulk of LC cell), cell viewing angle, and variations in cell thickness. The practical optimum contrast which is achieved in LC cells is in the range of 100:1 to 1000:1 [14]. Additionally, it has been shown that LC cells are practically immune to radiation damage up to at least 500 kGy which is adequate for even radiation therapy applications [15].

2.3. Amorphous Selenium Photoconductor

The ideal photoconductor should have sufficient x-ray absorption, high resolution, sensitivity, and low dark current. At present the only cost-effective material satisfying these requirements is a-Se. The clinical suitability of a-Se for medical x-ray imaging has been established previously in connection with xeroradiography [16] and it is also used in AMFPIs [2]. The x-ray sensitivity of a photoconductor depends on the x-ray absorption, the thickness of the detector, and the energy required to release an ehp. An attenuation of 50% is achieved with a 365 μm thick a-Se layer for a 50 keV beam and with a 30 μm thick layer for a 20 keV beam. For a bias field of 10 V/μm, ~ 400 ehp will be released from the absorption of at 20 keV photon and ~1000 ehp for a 50 keV photon.

Large area a-Se detectors are cost-effective and feasible because the photoconductor is amorphous and is made by evaporation techniques. They are easy scalable to the needed size without increased complexity or price. It also has a well established manufacturing process because of its extensive use in the photocopying industry.

2.4. The LC Cell

The LC cell used in the XLV is similar to the cells used in common LC displays. LC cell technology is well established and inexpensive LC cells can easily be made large enough to meet the requirements for general radiography.

2.4.1. Design criteria

There are many choices to be made in LC cell design. These include the materials used, the orientation of the LC molecules and polarizer(s), the way in which the molecules re-orient in response to the applied electric field, and the thickness of the cell. Cells may be designed to accommodate various application requirements: for example the wavelength of light to be used and range of electric field which will be applied.

For use as part of the XLV, we wish to design an LC cell which has an electro-optical characteristic curve sensitive to the small changes in the voltage across the cell caused by the charge accumulated on the a-Se-LC interface after exposure. The sensitivity must correspond to the amount of charge collected from x-ray exposures in range used in medical imaging. It is also desirable that the reflection as a function of exposure is monotonic and preferably linear. Response characteristics (e.g. linearity and operating range) are needed which will allow good contrast over the clinically relevant range of exposures. Different LC cells have varying optical characteristics [14] depending on both the configuration and the materials used. Hundreds of LC mixtures have been developed to meet different requirements. For an XLV operated in transmission mode, a design based on the 90° twisted nematic (TN) cell was proposed [9, 10]. The same design requires modifications to be used in reflective mode.

The image must be captured before the charge on the a-Se LC surface decays, so an important requirement for the XLV is a sufficient time constant τ [10]. $\tau \sim RC$ where R is the resistance of the LC mixture and C is the capacitance of the cell, which varies with the dielectric constant of the LC mixture. Since the dielectric constant of the LC materials available for purchase only varies over a small range, the parameter which is used to achieve a long τ is the resistance of the LC material. To maintain the image over a long time, a carefully chosen high resistivity or "high holding" LC material is used.

2.4.2. Characteristics of cells

To evaluate the performance of LC cells for use in the XLV, it is critical to determine the reflectance of each cell as a function of the amount of image charge collected on the surface of the a-Se, which in turn is proportional to the energy of the absorbed x-ray photons.

Figure 2(a) shows the reflectance-voltage characteristic curve, calculated for two example reflective TN (r-TN) cells by using LC modeling software, GNU-LCM [17]. The two examples shown are based on LC mixtures ZLI-4792 and E7, manufactured by Merck. The anisotropy in the index of refraction (Δn) is 0.0969 for ZLI-4792 and 0.2253 for E7. The corresponding cell gaps for 633 nm light were calculated to be 2.8 µm and 1.22 µm, respectively.

In our earlier work, the capacitance of the LC cell was assumed constant [9, 10], in which case the voltage is proportional to charge, $V = Q/C$, where V is the voltage across the LC, C is the capacitance of the LC cell, and Q is the charge created in the a-Se layer. However, LC cells exhibit a voltage dependent capacitance, C(V), due to the anisotropic dielectric constant of the rod-shaped LC molecules. The applied voltage causes the molecules to rotate, leading to the capacitance change. Figure 2(b) shows the capacitance change due to this effect. In this case, we have a non-linear relationship between charge and reflection. This is illustrated in Figure 2(c), where reflectance is plotted over the charge and corrected for the voltage dependant capacitance.

To evaluate the cells as part of an XLV, we make calculations using a monoenergetic beam of 50 keV, an energy commonly used in chest imaging. The number of photons per Roentgen is 2.842×10^{13} m^{-2}. Assuming an a-Se layer

500 µm thick (quantum efficiency of 57.1 %), a bias field of 10 V/µm, and 1000 ehp created per incident x-ray photon, the reflectance as a function of exposure is presented in Figure 2(d). This relationship can be used to predict the operation of XLVs with different cell designs and LC materials.

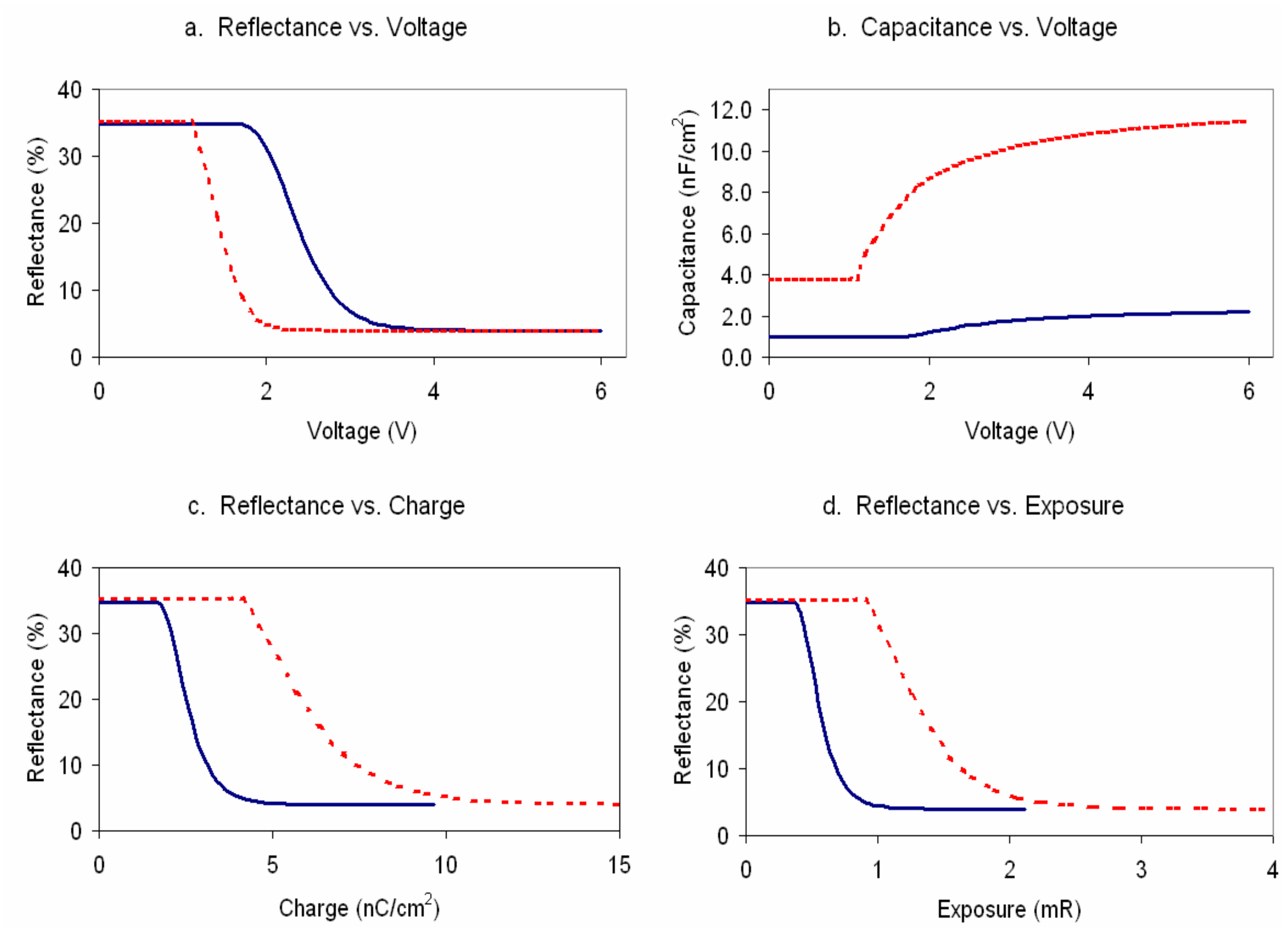

Figure 2: (a) Reflectance versus applied voltage for two example r-TN cells using different LC compounds: ZLI-4792 (solid lines) and E7 (dashed lines), (b) voltage dependent capacitance of both cells, (c) reflection versus collected charge, obtained by including the effect of the voltage dependent capacitance, and (d) reflectance versus exposure for both when incorporated in an XLV.

2.5. The Readout System

The latent charge image created on the surface of the a-Se layer upon interaction with x-rays creates variations in the reflectance of an LC cell. The cell modulates the intensity of incident light, which is reflected at the interface between the a-Se and the LC and is detected with a linear array of sensors. The light source and detectors are scanned across the XLV after exposure to acquire the visual image.

For the linear array of sensors, we use a Contact Image Sensor (CIS) head (see Figure 3). The light source used is a LED or multiple LEDs coupled to a light guide to create uniform illumination. The spatial resolution of these scanner heads is remarkably high – 600 dpi (42.3 µm pixels) or 1200 dpi (21.2 µm pixels), and even 4800 dpi (5.3 µm pixels) are available. This is comparable to the pixel size used in AMFPIs designed for digital mammography (40-50μm pixels). This visible and long-lived image contains the x-ray exposure information.

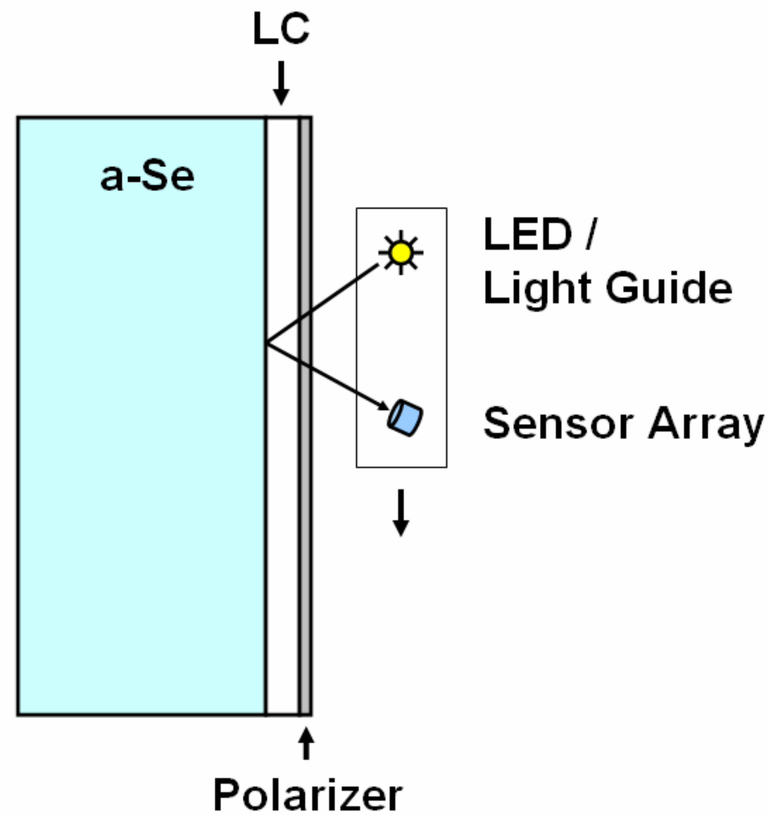

Figure 3: Readout of an XLV in reflective mode. A CIS is scanned across the XLV surface. A sheet polarizer placed between the XLV and the scanner head translates light-polarization changes in the LC layer to changes in light reflection.

The optical coupling in the readout need not cause the image degradation problems found in demagnified screen/camera systems because the XLV acts only to modulate the readout light intensity. The intensity of the readout light can be adjusted independently from the x-ray exposure to match the sensitivity of the readout system and to compensate for any loss of light in the optical path. Therefore, the x-ray image can be recorded with an optical imager without a secondary quantum sink [9].

The resolution of the XLV depends on both the inherent resolution of the photoconductor detector as well as the readout system. The Modulation Transfer Function (MTF) of the XLV has been previously derived in terms of blur in charge collection within the a-Se layer and the electric field spread in a two layer dielectric with an interface charge distribution [10]. The resolution limit in this process has been shown to be due to x-ray interactions (i.e. range of primary photoelectrons) [13]. The limiting element of the above described imaging chain was found to be the readout [10].

In Figure 4, we show the MTF of a Canon LiDE 30 optical scanner, which is sold as an inexpensive paper scanner in the under US$100 class. Its CIS scanner head is currently used in our readout system. Even with the limitations of the readout system, the XLV imaging chain should have adequate resolution for mammography applications, and since the resolution requirement for other imaging tasks (e.g. chest) is lower, it can be concluded that an XLV imaging system built using an optical scanner will have sufficient resolution for general radiography. By using a scanning approach and CIS as a readout system, we believe that the device can be scaled to the required size without sacrificing resolution, pixel density, or the low cost of the overall system. It will solve the problems stitching images from more than one area CCD as discussed in [11].

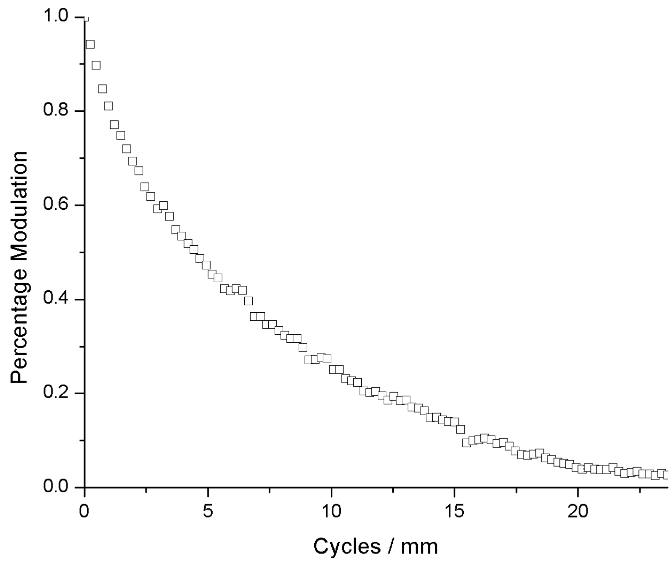

Figure 4: Modulation Transfer Function of a Canon LiDE 30 optical scanner.

3. SUMMARY

There is a need for a low cost imaging device for direct readout digital radiography systems. We have developed such a detector, the X-ray Light Valve. It is based on an a-Se photoconductor coupled to a liquid crystal cell. The image is read out with a modified commercial paper scanner. The XLV structure is simple – the photoconductor detector, the LC light modulator and the readout system are all well-established technologies. Little external circuitry is required beyond a conventional PC and the manufacturing costs could be kept very low. The detector size is easily scalable without sacrificing resolution, pixel density, or the low cost of the overall system. The XLV system has been shown to have all the properties required for general radiography.

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