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# NOVEL DETECTION TRENDS FOR DIGITAL MAMMOGRAPHY

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*Abstract: In this study, novel dual-energy detection modes for digital mammography, consisting of a combination of gas detector media with semiconductor detectors, are presented and discussed.*

*The purpose of this study is to measure experimentally the figure of merit (FOM) of this multimedia detector for dual-energy imaging. The experimental results indicate that the multimedia detector exhibits a high FOM. At present time, potential applications of the multimedia detector technology in several industrial areas, such as medical imaging, aerospace imaging, aviation security and surveillance, are being explored.*

*Keywords: digital mammography, gas detector media, multimedia detector*

## 1. INTRODUCTION

Extensive efforts over a number of years have been address to develop x-ray imaging sensors for digital radiography, radio-astronomy, x-ray crystallography, aviation security utilizing dual-energy line-scan sensors, x-ray computed tomography (CT) techniques, security x-ray equipment and non-destructive evaluation. The development of new x-ray sensors which will allow access to high image quality is of paramount significance.

The search for a suitable detection technique for dual-energy imaging is of paramount significance. The principles of dual-energy imaging [1]-[12] involve the use of two x-rays images, one produced from a high energy and another from a low energy polychromatic spectra. A weighted subtraction of these two images produces a digital image which eliminates interfering background structure.

Most current sensor approaches digital mammography employ a phosphor x-ray detector which, in response to x-ray absorption, produces light photons that are then converted into an electronic signal. This process is inefficient and can lead to increased image noise, particularly when signals are low. Our approach is to avoid the problems associated with the phosphor type detectors by using direct conversion solid state detectors. In these detectors the x-ray photon directly creates electron-hole pairs without producing the intermediate light photons. Semiconductor detector arrays are potentially attractive because of their direct conversion of x-rays to electrical signals, in a variety of imaging applications with emphasis in aerospace and medical industry.

Specifically, the use of high density, high Z detector media such as CdZnTe semiconductors, combined with an efficient collector design would allow for a decreased detector thickness, with a high collector efficiency, and consequently improved spatial and contrast resolution. As a result of the proposed sensor technology, an enhanced image quality will result. To highlight the potential of CdZnTe detectors, us assume x-ray photons, with an energy of 20 keV, typical for low-energy x-ray imaging applications such as mammography, incident on a 1mm thick CdZnTe substrate. As a result, 4514 $e^-$/holes pairs will be created. The quantum efficiency at the same physical conditions is greater >99%. Therefore, a good SNR is produced. Specifically, assuming a typical mammographic exposure per frame of 10 mR, with a typical exposure time of 1 s, and a conversion factor of $5.6 \times 10^7$ photons/mm$^2$/R, a photon fluence of $5.6 \times 10^5$ photons/mm$^2$ or an exposure/pixel of 1390 photons/pixel, results. Based on the above, the associated charge/pixel is equal to $2.2 \times 10^{-16}$ Coulombs, the Poisson noise equals 37.2 photons/pixel , the Poisson noise level/pixel is of the order of 2.6% and the signal-to-noise ratio is 37.2. Based on the Rose model, images with a SNR between 3-5 or higher can be considered detectable. A readout noise less than 25,000 electrons RMS, corresponds to less than 5.5 incident photons/pixel. Therefore, the detection process is quantum limited.

Recently, the principles of a multimedia detector for medical imaging, operating on gaseous solid state ionization principles, with specific emphasis on single x-ray exposure dual-energy radiography, have been introduced and presented [13]-[18]. A schematic diagram of the multimedia detector is shown in Fig. 1.

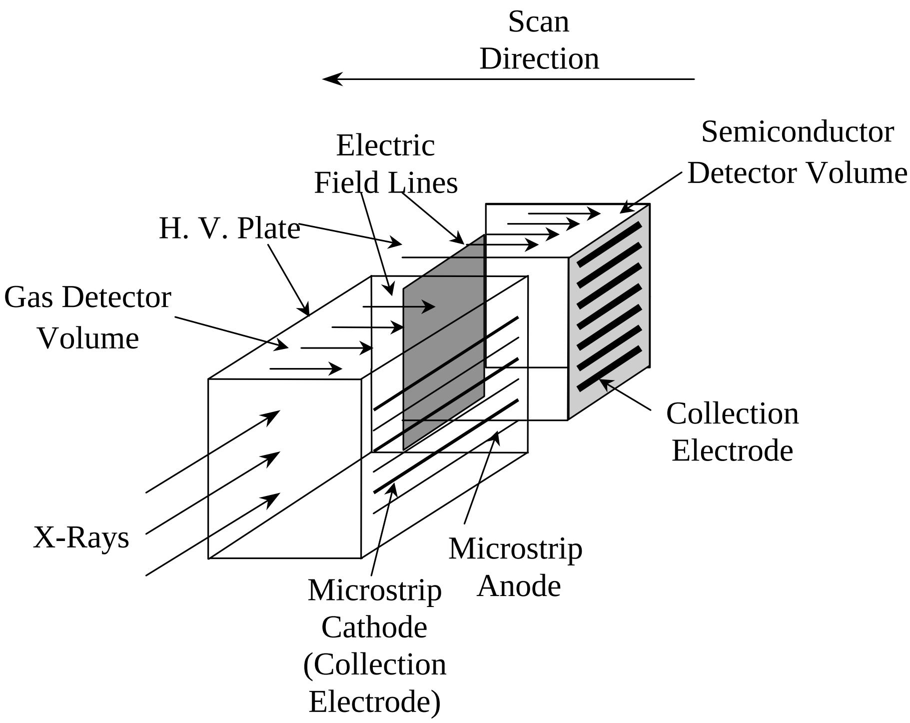


**Figure 1.** The multimedia detector for dual-energy imaging

Specifically, the multimedia detector consists of two detector elements, namely, a front detector element and a rear detector element, separated by an inactive midfilter segment. The front detector element, a gas microstrip detector [19]-[21], produces the digital low-energy photon image, and the rear detector element, a semiconductor detector, i.e., $Cd_{1-x}Zn_xTe$ [22]-[23], produces the digital high-energy photon image. One advantage of the multimedia detector technology is a large signal amplification in the front detector element, due to the intrinsic properties of the gas microstrip. As a result, a low radiation dose may be utilized at relatively low operating gas pressures. This detector technology may also give superb spatial resolution due to 1) the ultrafine structure of the microstrip substrate, 2) the high scatter rejection of the slot-scanning beam detector geometry, 3) reduced space-charge effects due to the rapid ion collection time, and 4) high versatility arising from the optimization of the front detector element upon suitably chosen operating gas pressure and gas mixture. On the other hand, a high absorption efficiency may result by utilizing a high density (5.8 g/cm$^3$ ), high effective atomic number (Z=46) CdZnTe semiconductor substrate as the rear detector element [22]-[23]. It is well known, that high quality CdZnTe semiconductor crystals have been grown using the High-Pressure Bridgman (HPB) technique. Specifically, by alloying CdTe with Zn, the bulk resistivity of this semiconductor is approximately $10^{11}$ $\Omega$-cm. This high resistivity is due to the wide band gap of this ternary semiconductor which results in low leakage currents and consequently, low noise characteristics.

## 2. EXPERIMENTAL RESULTS AND DISCUSSION

The x-ray generator used in this study was a three-phase Picker 612 which powered a Dunlee PX-1842-AQ x-ray tube, with a 0.6 mm focal spot. The focal spot-to-detector distance was 1 m. The "dead" volume, i.e., the x-ray detection volume which does not contribute electrons to the gas-strips, extends 1.2 cm into the xenon volume. Initially, the test detector was flushed with nitrogen for 10 hours. Incident x-rays spend part of their energy in the xenon gas detector volume, (for an active depth of interaction of 1 cm), and the other part through the interaction in the solid state detector volume, producing ion-electron pairs and holes-electrons pairs, respectively. An applied electric field imparts a constant drift velocity to these charges, driving them towards their respective signal collectors. The detected charge signal was amplified by an AMTEK 250 charge sensitive preamplifier. To allow signal optimization of the electronic system, a 2SK152 field effect transistor (FET) with a small input capacitance was connected to the input of the charge sensitive preamplifier, so that it could be matched to the low capacitance of the $Cd_{1-x}Zn_xTe$ detector, as well as for noise and shaping requirements. The noise characteristics of the preamplifier as a function of the detector capacitance are such that its contribution to FET and detector noise is negligible, specifically, between 120 to 130 electrons RMS for detector capacitances between 1 pF and 10 pF. A 2 pF feedback capacitance was set in parallel to a 100 MΩ feedback resistance. The output signal was then filtered by a low-pass fourth-order active Butterworth filter. The 3-dB roll-off frequency of the system was measured to be approximately 20 kHz. The total amplification gain was 1. The signal was then digitized at 200K samples/s through a National AT-MI0-16E-1 12 bit A/D converter, then stored and displayed on a PC monitor. In order to assess the sensitometric response of the multimedia detector, preliminary signal-to-noise measurements were performed with varying gas microstrip anode bias voltage, tube current setting, and incident photon energy. Then noise data were obtained by recording three air-scans. In each air-scan set of measurement, a uniform (except for statistical fluctuations) portion of waveform signal was isolated, consisting of 50 data points within the same range of observation. In each of the three cases, the standard deviation from the mean signal value was estimated. The signal-to-noise ratio in each case was computed by dividing the signal by the standard deviation, and an average signal-to-noise ratio was calculated.

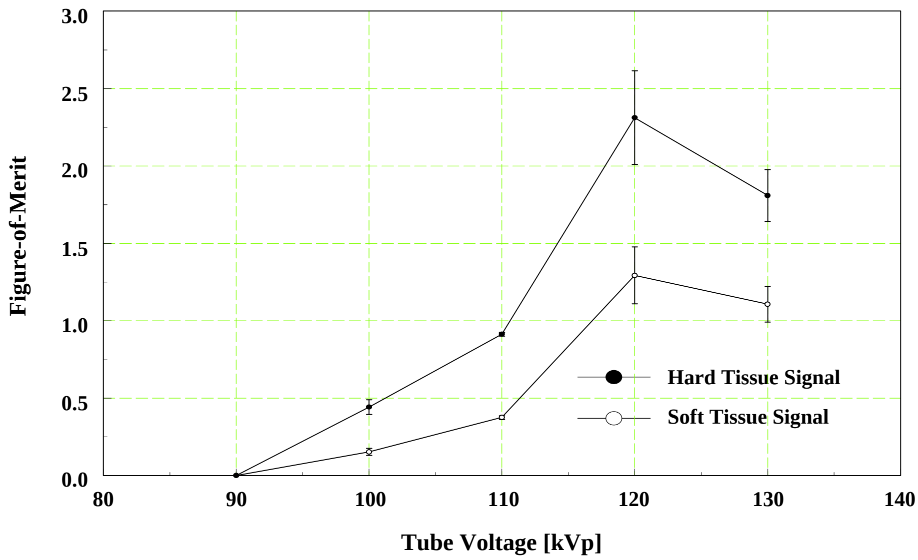


**Figure 2.** Hard and soft tissue figure-of-merits versus tube voltage for a 55 mm Plexiglas and 0.4 mm aluminum phantom, using 100 mA, 1680 V MS bias and 75 V/mm CdZnTe bias.

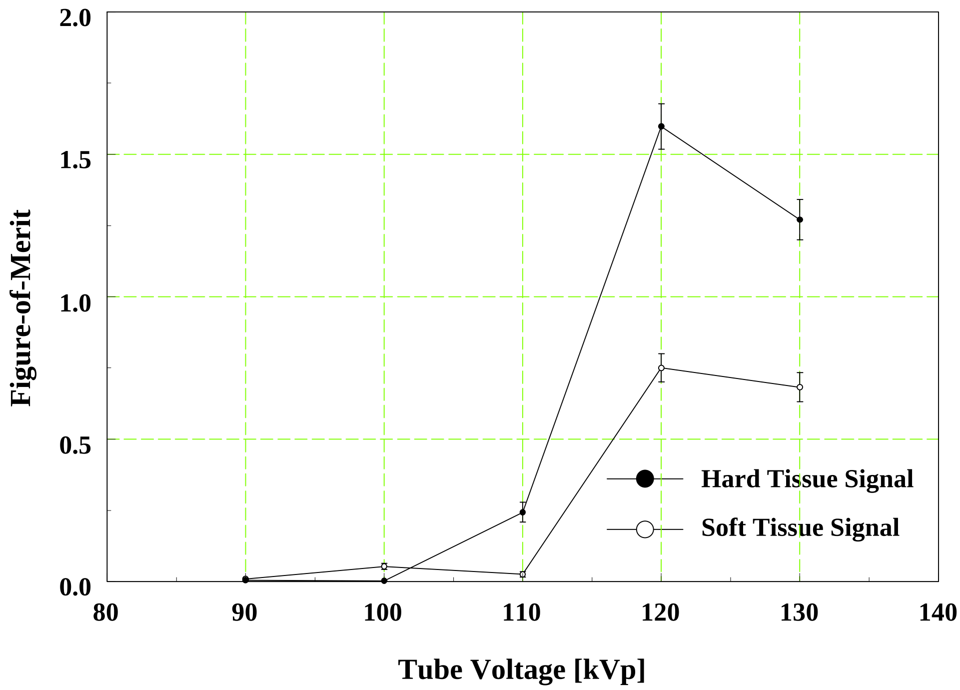


**Figure 3.** Hard and soft tissue figure-of-merits versus tube voltage for a 55 mm Plexiglas and 2 mm aluminum phantom, using 100 mA, 1696 V MS bias and 75 V/mm CdZnTe bias.

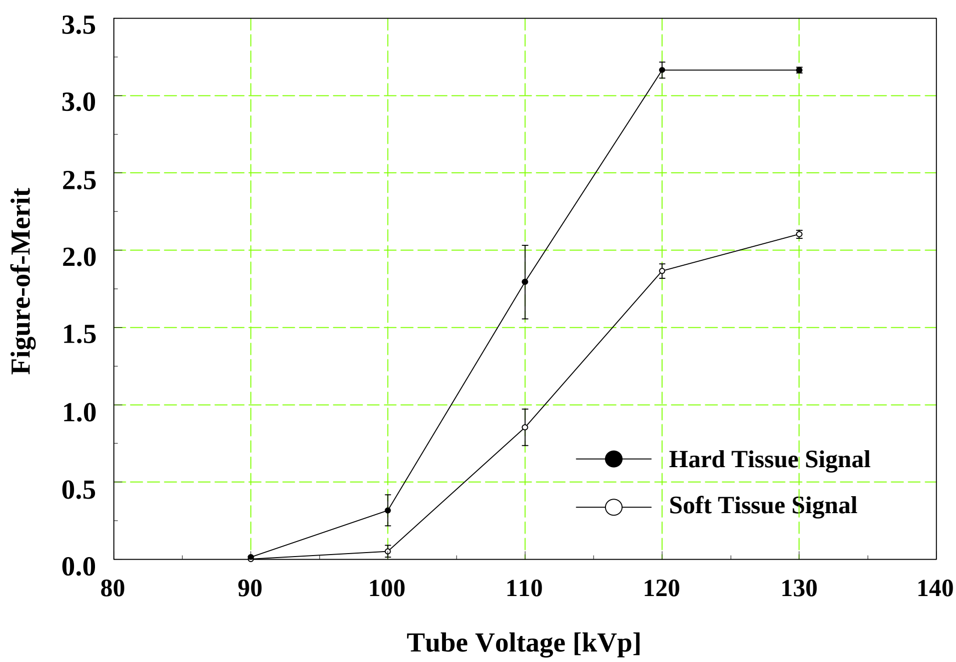


**Figure 4.** Hard and soft tissue figure-of-merits versus tube voltage for a 150 mm Plexiglas and 0.4 mm aluminum phantom, using 100 mA, 1692 V MS bias and 100 V/mm CdZnTe bias.

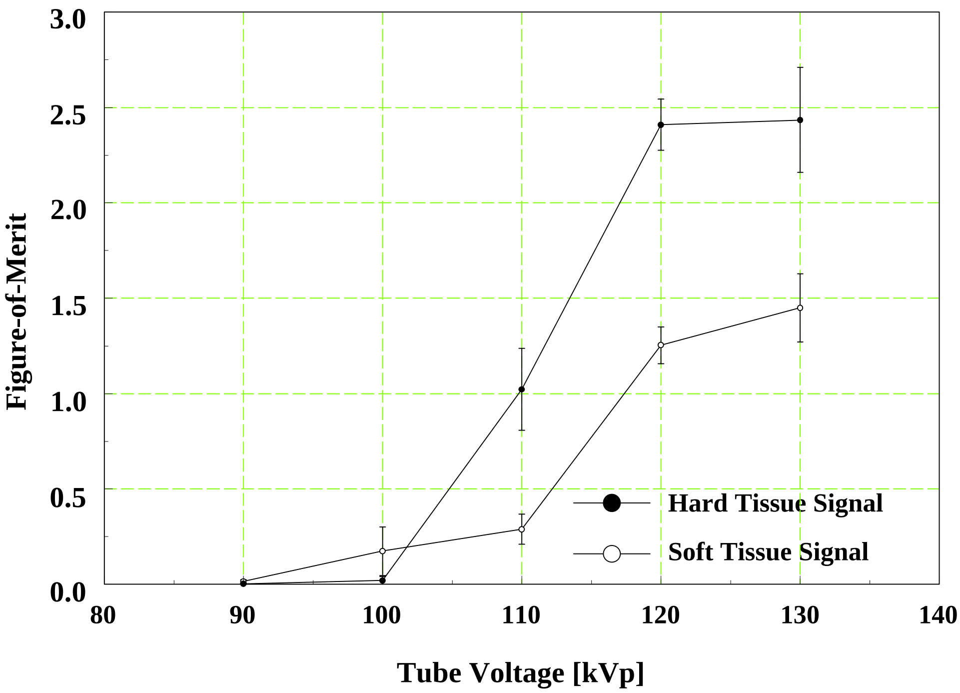


**Figure 5.** Hard and soft tissue figure-of-merits versus tube voltage for a 150 mm Plexiglas and 2 mm aluminum phantom, using 100 mA, 1698 V MS bias and 100 V/mm CdZnTe bias.

The experimental results of this study, as shown in Fig. [2]-[5]. In these figures, the hard and soft tissues FOMs versus tube voltage for each of the four phantom configurations, are shown.

The FOMs increased significantly with increasing tube voltage. This can be attributed to a combination of factors. For instance, higher energy photons lead to a decreased interaction with the phantom, a higher fluence incident on the detector, and a higher SNR. However, a more important factor involves the energy spectrum of the photons produced by the x-ray tube. By increasing the tube voltage, the photon energy spectrum shifts up leading to a higher effective energy absorbed by the rear detector, yielding a greater energy separation. However, in opposition to these factors, increasing the tube voltage will increase the dose and eventually cause a loss in image information. These last two effects place an upper limit on the tube voltage as can be observed in Fig. 2 and 3 where the FOM peaks at 120 kVp then decreases. In Fig.s 4 and 5, using the thicker 2 mm aluminum phantom tube voltages of 120 kVp and 130 kVp provided the highest FOMs.

## 3. CONCLUSION

The experimental results of this study, clearly indicate that the multimedia detector exhibits superior dual-energy capabilities. Overall, it is believed that the presented technology is of paramount significance for several imaging areas as applied to medical detector technology, space research, for the design of high efficiency detectors for radio-astronomy, aviation security, surveillance and non-destructive inspection, and other industrial applications.

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