August 2001

Signal-to-noise measurements utilizing a novel dual-energy multimedia detector

G. C. Giakos
The University of Akron

S. Chowdhury
The University of Akron

N. Shah
The University of Akron

Srinivasan Vedantham
The University of Akron

A. G. Passerini
University of Pennsylvania

See next page for additional authors

Signal-to-noise measurements utilizing a novel dual-energy multimedia detector

Abstract

Dual-energy measurements are presented utilizing a novel slot-scan digital radiographic imaging detector, operating on gaseous solid state ionization principles. The novel multimedia detector has two basic functional components: a noble gas-filled detector volume operating on gas microstrip principles, and a solid state detector volume. The purpose of this study is to investigate the potential use of this multimedia detector for enhanced dual-energy imaging. The experimental results indicate that the multimedia detector exhibits a large subtracted signal-to-noise ratio. Although the intrinsic merit of this device is being explored for medical imaging, potential applications of the multimedia detector technology in other industrial areas, such as aerospace imaging, aviation security, and surveillance, are also very promising.

Keywords

radiographic imaging, signal-to-noise ratio, x-ray energy spectrum

Comments

Author(s)

G. C. Giakos, S. Chowdhury, N. Shah, Srinivasan Vedantham, A. G. Passerini, Sankararaman Suryanarayanan, R. Nemer, K. Mehta, N. Patnekar, S. Sumrain, K. Nataraj, and Christian Scheiber

Signal-to-Noise Measurements Utilizing a Novel Dual-Energy Multimedia Detector

G. C. Giakos, *Senior Member, IEEE*, S. Chowdhury, *Student Member, IEEE*, N. Shah, Srinivasan Vedantham, *Member, IEEE*, A. G. Passerini, Sankararaman Suryanarayanan, *Member, IEEE*, R. Nemer, K. Mehta, N. Patnekar, S. Sumrain, K. Nataraj, and Christian Scheiber

***Abstract*—Dual-energy measurements are presented utilizing a novel slot-scan digital radiographic imaging detector, operating on gaseous solid state ionization principles. The novel multimedia detector has two basic functional components: a noble gas-filled detector volume operating on gas microstrip principles, and a solid state detector volume. The purpose of this study is to investigate the potential use of this multimedia detector for enhanced dual-energy imaging. The experimental results indicate that the multimedia detector exhibits a large subtracted signal-to-noise ratio. Although the intrinsic merit of this device is being explored for medical imaging, potential applications of the multimedia detector technology in other industrial areas, such as aerospace imaging, aviation security, and surveillance, are also very promising.**

***Index Terms*—Radiographic imaging, signal-to-noise ratio, X-ray energy spectrum.**

I. Introduction

THE search for a suitable detector for dual-energy digital radiographic imaging is of paramount significance. The principles of dual-energy imaging involve the use of two x-ray images, one produced from a high energy and another from a low energy polychromatic spectrum. A weighted subtraction of these two images produces a digital image that eliminates interfering background structures. By simplifying the background structure in this way, an increase in the detectability or "conspicuity" of the target structure is obtained.

Dual-energy imaging has been explored by several investigators [1]–[12] with the main emphasis on mammography and chest radiography. Dual-energy digital chest radiography permits cancellation of the displayed contrast of any two materials, such as bone and soft tissue, allowing low-contrast lesions, such as solitary pulmonary nodules or diffused pulmonary nodules to be obtained through increased detectability in bone-canceled images. Similarly, digital dual-energy mammography enhances the detection of calcifications by removing structural noise from the surrounding soft tissue in the breast. It may, therefore, be viewed as an attempt to maximize the diagnostic information content of the x-ray image.

Dual-energy radiographic detection is performed utilizing either a single x-ray exposure, which produces high and low photon energy images simultaneously on a split detector system, or through a double x-ray exposure, which produces high and low energy images separately on a single detector element. In the single exposure technique, two different detector elements are utilized consisting of a low energy front detector and a high energy rear detector. A suitably chosen prefilter is placed in front of the poly-energetic x-ray beam, forming a bimodal polychromatic spectrum, with peaks separated at low and high energies. The front detector absorbs most of the low energy photons, while the rear detector absorbs most of the high-energy photons. The advantages of this technique are that it eliminates misregistration problems and patient motion artifacts, which otherwise may be present in the double x-ray exposure dual-energy technique. However, signal reduction may occur, either because of the use of the prefilter, which absorbs part of the x-ray energy spectrum, or because of the partial absorption of the high-energy photons in the front detector. Therefore, a reduced signal-to-noise ratio in the rear detector may result.

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. 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., $\mathrm{Cd}_{1-x}\mathrm{Zn}_x\mathrm{Te}$ [22], [23], produces the digital high-energy photon image. One advantage of the multimedia detector technology is 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) ultrafine structure of the microstrip substrate;
2) high scatter rejection of the slot-scanning beam detector geometry;
3) reduced space-charge effects due to the rapid ion collection time;
4) high versatility.

Manuscript received May 4, 2000; revised April 2, 2001. This work was supported in part by an Institute for Biomedical Engineering Research (IBER) Faculty Enhancement Grant.

G. C. Giakos, S. Chowdhury, N. Shah, S. Vedantham, A. G. Passerini, S. Suryanarayanan, R. Nemer, K. Mehta, N. Patnekar, S. Sumrain, and K. Nataraj are with the Olson Research Center, Imaging Devices, Sensors and Communications Laboratory, Department of Biomedical Engineering, The University of Akron, Akron OH, 44325 USA.

C. Scheiber is with the Institut de Physique Biologique, Strasbourg Cedex, France.

Publisher Item Identifier S 0018-9456(01)06024-7.

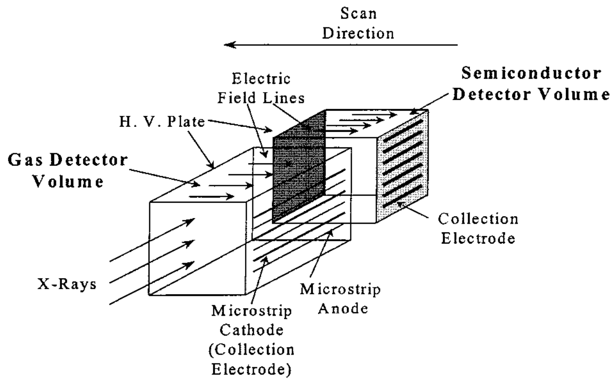

Fig. 1. Multimedia dual-energy detector proposed by Giakos [13]–[16].

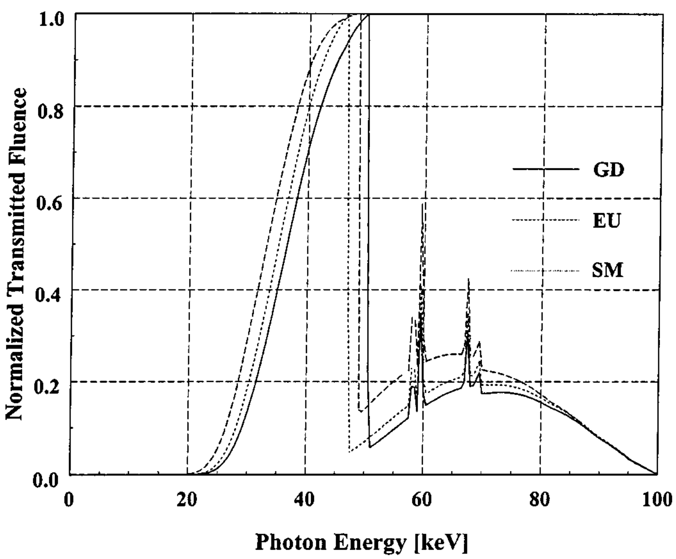

Fig. 2. Transmitted fluence spectrums through 0.25 mm Gadolinium (Gd), Europium (Eu), and Samarium (SM) prefilters, at 100 kV$_p$.

The last reason arises from 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 technique. Specifically, by alloying CdTe with Zn, the bulk resistivity of this semiconductor is approximately 10^{11} Ω-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.

II. Prefilter Selection

Simulation software was used to determine the best choice of the prefilter. Materials with k-edges in the range of (40–50) keV, such as Barium (Ba), Cerium (Ce), Europeum (Eu), Gadolinium (Gd), Iodine (I), Lanthanum (La), Molybdenum (Mo), Samarium (Sm), Tellurium (Te), and Neodymium (Nd), were examined. The k-edge is the energy at which a local maximum photoelectric absorption occurs and is determined

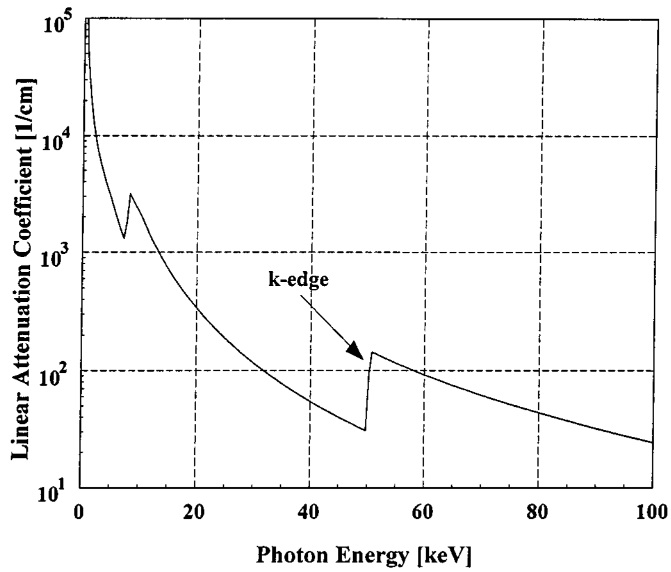

Fig. 3. Linear attenuation coefficient versus photon energy for Gadolinium.

by the binding energy of the k-shell electrons. Representative spectra transmitted by different prefilter materials are shown in Fig. 2. Dual-energy simulations were run with varying tube voltages, prefilter type, and thickness. Several phantom cases representing mammography, chest radiography, and osteoporosis were used. Detector parameters such as detector element thickness and gas pressure were also varied. The figure-of-merit in each case was estimated. Gd and Sm prefilters consistently produced the best results. In these experiments, a Gd prefilter has been utilized to transform the x-ray spectrum into a bimodal one. The linear attenuation coefficient of a Gd prefilter is shown in Fig. 3.

The purpose of this study is to investigate the potential use of this multimedia detector for enhanced dual-energy imaging. The experimental results indicate that the multimedia detector exhibits a high sensitivity [15]. Although the intrinsic merit of this device is being explored for medical imaging, potential applications in other industrial areas, such as aerospace imaging, are also very promising.

III. Experimental Arrangement and Results

The multimedia detector has two basic functional components: a gas filled detector volume consisting of a gas microstrip detector and a solid state detector volume consisting of a planar $Cd_{1-x}Zn_xTe$ sample, operating under scanning beam geometry (Fig. 1). The whole system is enclosed in a vessel made of high tensile strength aluminum alloy. The front window is 1 mm thick and composed of the same aluminum alloy as the test detector. Field shaping electrodes were placed in the proximity of the detector window. In order to test the sensitometric response of the test detector, a collimator consisting of a 1.5 cm $\times$ 10 cm lead sheet, 2 mm thick, with a slit aperture of 1 mm, was set 5 cm in front of the detector window. The collimation in the present study extends 1 mm into the drift region from the microstrip substrate. The microstrip substrate (12.5 cm $\times$ 12.5 cm) consists of 25 anode strips coated with 3000 Å of chromium, interleaved with cathode strips. The electric potential alternates between

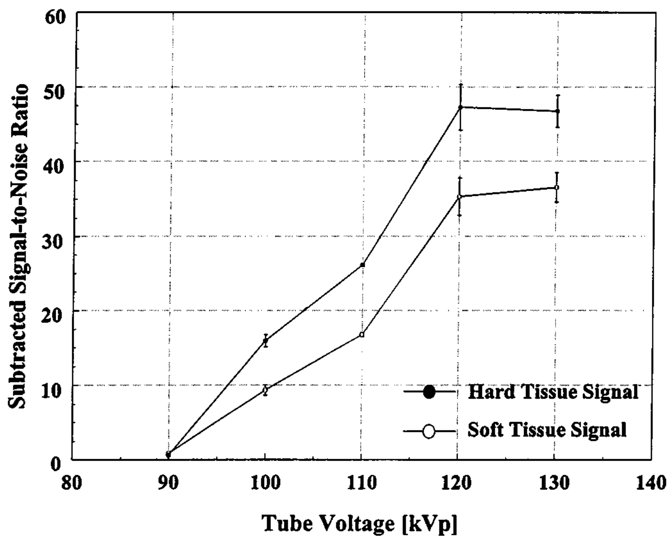

Fig. 4. Hard and soft tissue subtracted signal-to-noise ratios versus tube voltage for a 55 mm Plexiglas and 0.4 mm aluminum phantom, using 100 mA, 1680 V RMS bias and 75 V/mm CdZnTe bias.

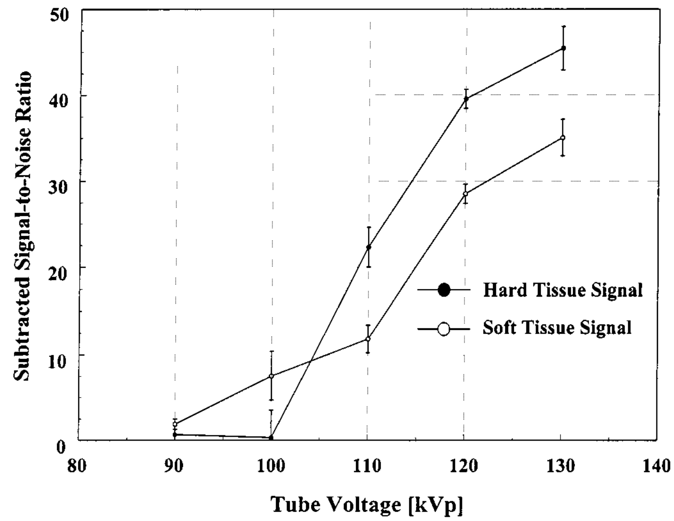

Fig. 5. Hard and soft tissue subtracted signal-to-noise ratios versus tube voltage for a 55 mm Plexiglas and 2 mm aluminum phantom, using 100 mA, 1696 V RMS bias and 75 V/mm CdZnTe bias.

each strip. The whole system is adhered on a Corning 7740 substrate. The microstrip substrate has been fabricated with 25 μn wide anode strips, and 625 μn wide cathode strips. It should be noted that these collectors are much wider than a typical detector would be, i.e., in the order of (40–100) micron-pitch width. Xenon gas was used in these experiments because of its high x-ray stopping power. The distance between the high voltage plate and the microstrip substrate, i.e., the drift gap, is approximately 10 mm. The high voltage plate and the microstrip signal collectors form the two main electrodes of the proposed technology. The CdZnTe detector design is planar, 5 mm $\times$ 5 mm, with a thickness of 1 mm, with gold electrodes applied by the electroless method on opposite faces providing an electric field across the semiconductor detector for an applied bias voltage. In the experiments described in this paper, the x-ray beam is incident on the negative electrode, which also serves as the collector, with the electric field parallel to the direction of the incoming photons.

The x-ray generator used in this study was a three-phase Picker model 612, which powered a Dunlee model 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 h 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-electron pairs, respectively. An applied electric field imparts a constant drift velocity to these charges, driving them toward their respective signal collectors. The detected charge signal was amplified by an AMTEK model 250 charge sensitive preamplifier. To allow signal optimization of the electronic system, a type 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

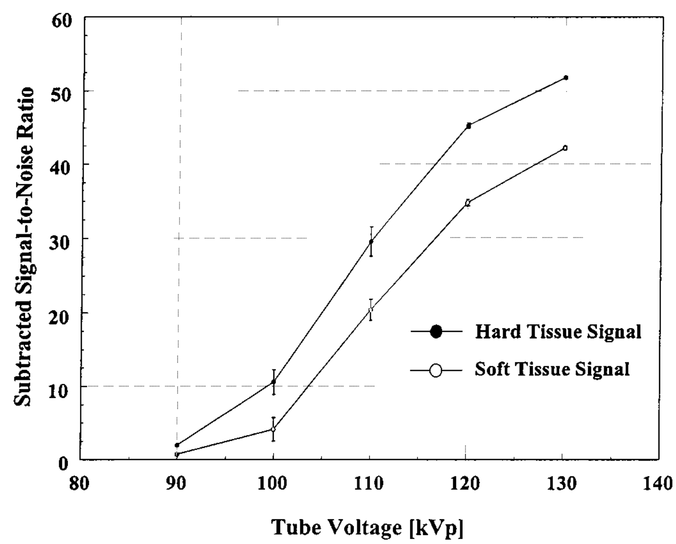

Fig. 6. Hard and soft tissue subtracted signal-to-noise ratios versus tube voltage for a 150 mm Plexiglas and 0.4 mm aluminum phantom, using 100 mA, 1692 V RMS bias and 100 V/mm CdZnTe bias.

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. Å 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 200 ksamples/s through a National model 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

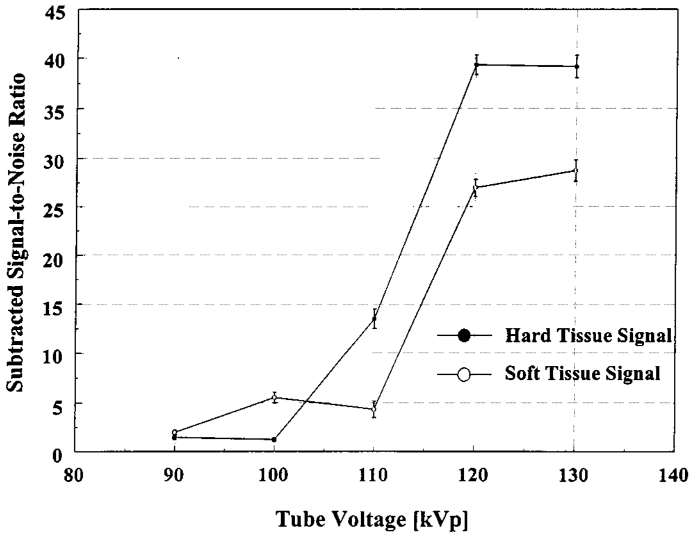

Fig. 7. Hard and soft tissue subtracted signal-to-noise ratios versus tube voltage for a 150 mm Plexiglas and 2 mm aluminum phantom, using 100 mA, 1698 V RMS bias and 100 V/mm CdZnTe bias.

were obtained by recording three air-scans. In each air-scan set of measurement, a uniform (except for statistical fluctuations) portion of the 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.

In Figs. 4–7, the subtracted signal-to-noise ratios versus tube voltage for both hard and soft tissue signals are shown for different phantom configurations. It is observed that the subtracted signal-to-noise ratio increases with increasing x-ray tube voltage. This is attributed to the fact that the mean energy of the x-ray spectrum shifts up, increasing the number of the high- energy photons. Therefore, higher energy fluence on the detector and a higher signal-to-noise ratio result. Furthermore, at reduced Plexigas phantom thickness (55 mm), the subtracted signal saturates at high x-ray tube voltages (Figs. 4 and 5). This effect is not apparent in Figs. 6 and 7, where a thicker (150 mm) Plexiglas phantom is used.

IV. Conclusion

The results of this experimental study indicate that the multimedia detector exhibits a high subtracted signal-to-noise ratio. Overall, the multimedia detector shows a great potential for use in dual-energy digital radiography and other x-ray imaging applications.

References

[1] H. N. Cardinal and A. Fenster, "Theoretical optimization of a split septaless xenon ionization detector for dual-energy chest radiography," *Med. Phys.*, pp. 167–180, 1988.
[2] W. R. Brody *et al.*, "A method for selective tissue and bone visualization using dual energy scanned projection radiography," *Med. Phys.*, vol. 8, p. 353, 1981.
[3] L. A. Lehman *et al.*, "Generalized image combinations in dual KVP digital radiography," *Med. Phys.*, vol. 8, p. 659, 1981.
[4] C. E. Cann *et al.*, "Quantification of calcium in solitary pulmonary nodules using single-and dual energy CT," *Radiology*, vol. 145, p. 493, 1982.
[5] D. P. Chakraborty and G. T. Barnes, "An energy sensitive cassette for dual-energy mammography," *Med. Phys.*, vol. 16, pp. 7–13, 1989.
[6] W. R. Brody *et al.*, "Dual-energy projection radiography: Initial clinical experience," *Am. J. Roentgenol*, vol. 137, pp. 201–205.
[7] J. M. Boone, G. S. Shaber, and M. Tecotzky, "Dual-energy mammography: A detector analysis," *Med. Phys.*, vol. 17, pp. 665–675, 1990.
[8] H. L. Kundel *et al.*, "Nodule detection with and without a chest image," *Invest. Radiol.*, vol. 20, p. 94, 1984.
[9] R. J. Endorf, S. Kulatunga, D. C. Spelic, S. R. Thomas, F. A. DiBianca, G. C. Giakos, and H. D. Zeman, "Clinical applications of a dual energy KCD," *Proc. SPIE*, vol. 2132, pp. 222–233, 1994.
[10] R. J. Endorf, S. Kulatunga, D. C. Spelic, S. R. Thomas, F. A. DiBianca, H. D. Zeman, and G. C. Giakos, "Preliminary performance characteristics of a dual-energy KCD," *Proc. SPIE*, vol. 2432, pp. 607–615, 1995.
[11] A. Dasgupta, "A Dual Energy Cadmium Zinc Telluride Telluride Detector System for Digital Radiography," M.S. thesis, Univ. Akron, Akron, OH, 1998.
[12] R. J. Endorf, S. Kulatunga, D. C. Spelic, S. R. Thomas, F. A. DiBianca, H. D. Zeman, and G. C. Giakos, "Preliminary performance characteristics of a dual-energy KCD," *Proc. SPIE*, vol. 2432, pp. 607–615, 1995.
[13] G. C. Giakos, *Hybrid Detection Trends in Medical Imaging*, ser. Physics Seminar Series. Ottawa, ON, Canada: Herzberg Laboratory of Physics, Carleton Univ., May 1997.
[14] ——, "A slot-scanned detector operating on gas-solid state imaging principles," *Proc. IEEE IMTC*, vol. 1, pp. 352–357, May 18–21 1998.
[15] ——, "Key paradigms of emerging imaging sensor technologies," *IEEE Trans. Instrum. Meas*, vol. 40, pp. 1–9, Dec. 1998.
[16] ——, "Emerging imaging sensor technologies: From aerospace to healthcare," *IEEE Instrum. Meas. Mag.*, pp. 16–19, Dec. 1998.
[17] G. C. Giakos and S. Chowdhury, "Multimedia imaging detectors operating on gas-solid state ionization principles," *IEEE Trans. Instrum. Meas.*, vol. 40, pp. 1–9, Oct. 1998.
[18] S. Chowdhury, "Experimental study of a novel multimedia sensor for dual-energy imaging," Ph.D. dissertation, Univ. Akron, Akron, OH, Dec. 1999.
[19] A. Oed, "Position-sensitive detector with microstrip anode for electron multiplication with gases," *Nucl. Instrum. Meth. Phys. Res. A*, vol. 263, pp. 351–359, 1988.
[20] G. C. Giakos, S. Vedantham, S. Chowdhury, and B. Pillai, "Novel hybrid imaging modalities," in *Proc. SPIE*, 1997, pp. 476–476.
[21] S. Chowdhury, "Signal Production of gas-microstrip detectors for medical imaging," M.S. thesis, Univ. Akron, Akron, OH, 1995.
[22] S. Vedantham, "Assessment of cadmium zinc telluride for x-ray imaging," M.S. thesis, Univ. Akron, Akron, OH, 1997.
[23] G. C. Giakos, A. Dasgupta, S. Suryanarayanan, S. Chowdhury, R. Guntupalli, S. Vedantham, B. Pillai, and A. Passalaqua, "Sensitometric response of CdZnTe detectors for chest radiography," *IEEE Trans. Instrum. Meas.*, vol. 47, pp. 252–255, Feb. 1998.

G. C. Giakos (M'92–SM'97) joined the Faculty of the University of Akron, Akron, OH, in 1994. His primary research interests are in the design, development, and experimentation of cost-effective, high-resolution imaging technologies, detectors, and sensors. In addition, he heads the Imaging Devices, Detectors, and Sensors Laboratory, which provides students with in-depth knowledge in multidisciplinary areas of imaging research, such as medical detector technology, plasma electronics, semiconductor devices, digital radiography and computer tomography, and magnetic tomographic imaging. He has reseach interests in the areas of smart sensors and transducers, RF and microwaves, and lasers and lightwave applications. He has published more than 100 papers in different areas of science adn technology. He is a member of the Editorial Board of the *Biomedical Instrumentation and Technology Journal*.

Dr. Giakos is National Chairman of the IEEE Instrumentation and Measurements Technical Committee on Imaging Systems. He is an Associate Editor of the IEEE Transactions on Instrumentation and Measurement.

S. Chowdhury (S'01) received the B.S.E.E. degree from the university of Michigan, Ann Arbor, in 1993 and the M.S. degree in biomedical engineering from the University of Akron, Akron, OH, in 1996, where he is currently pursuing the Ph.D. degree in imaging systems for medical applications.

His master's work involved optimization of gas microstrip detectors for digital radiography. He has ten publications in technical journals and conference proceedings.

N. Shah received the B.E. degree in biomedical engineering from the University of Bombay, Bombay, India, in 1998. She is currently pursuing the M.S. degree in biomedical engineering at the University of Akron, Akron, OH.

She is a Research Assistant with the Medical Imaging Devices and Sensors Laboratory, University of Akron. Her current research interests are related to the design of novel imaging technologies with improved resolution and reduced dose.

Srinivasan Vedantham (M'00) received the undergraduate degree in electrical engineering from the Birla Institute of Technology and Science, India, and the M.S. degree in biomedical engineering from the University of Akron, Akron, OH.

He joined the Radiologic Physics Research Group at the University of Massachusetts Medical Center, Worcester, in 1997. He also worked with the medical engineering division of Siemens, India. His primary interest is in the development, characterization, and optimization of digital x-ray detectors for radiography, fluoroscopy, and mammography.

A. G. Passerini, photograph and biography not available at the time of publication.

Sankararaman Suryanarayanan (M'99) received the undergraduate degree in electronics and communications engineering from the University of Madras, India, and the M.S. degree in biomedical engineering from the University of Akron, Akron, Ohio.

In 1998, he joined the Radiologic Physics Research group at the University of Massachusetts Medical Center, Worcester, as a Research Engineer. His primary field of interest is in digital imaging system development, simulation, modeling, and characterization. He currently works on digital imaging applications including tomosynthesis, mammography, fluoroscopy, and radiography.

R. Nemer was born in Akron, OH, in 1957. He received the B.S. degree in electronic engineering technology with a minor in instrumentation technology from the University of Akron in 1983.

From 1976 to 1978, he was with Radian Electronics, Akron, as an Electronics Designer and Special Project Coordinator. From 1979 to 1982, he was with the Departments of Biology and Chemical Engineering at the University of Akron. Since 1983, he has been with the Department of Biomedical Engineering and the Institute for Biomedical Engineering Research, College of Engineering, University of Akron.

K. Mehta, photograph and biography not available at the time of publication.

N. Patnekar, photograph and biography not available at the time of publication.

S. Sumrain, photograph and biography not available at the time of publication.

K. Nataraj, photograph and biography not available at the time of publication.

Christian Scheiber, photograph and biography not available at the time of publication.