A field emission x-ray tube for mamography

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ABSTRACT

We propose to develop a time-gated x-ray imaging technique for screening mammography. This concept challenges the current techniques in scatter rejection x-ray mammography. If successful, it will greatly improve the accuracy of current mammography with about 50% reduction of current patient radiation. The technique can be applied to other x-ray imaging procedures also. We are now working to determine the feasibility of the key technologies—an x-ray source that can be pulsed on and off at a subnanosecond rate and a detector that can be gated at the same rate—that represent potential roadblocks to achieving this goal. A field emission x-ray tube will be used as the source. The field emission cathode consists of a substrate with sharp points and a metallic grid. A voltage on the grid creates an electric field that produces tunnel emission from the array of tips. The geometry of such a tube make electron optics easier, high currents are possible and the tube can be turned on and off rapidly. It also appears that synchronous electronic gating of the detector is possible.

1. Introduction

It is well known that scattered radiation degrades the contrast of x-ray images in diagnostic radiology. According to measurements made by Wagner, scatter to primary ratio (S/P) in mammography typically ranges from 0.3 to 1.0 [1991]. As a result, subject contrast is degraded. Only 50% to 70% of the possible (primary beam) contrast is imaged in a clinical mammogram [1].

In order to minimize this problem, current practice is to use a Bucky grid between the subject and the detector. In this manner, x-ray image contrast may be improved by 10% to 40%[2]. Unfortunately, the grid also absorbs some of the primary radiation as well. This may result in a 2-3 times greater radiation dose to the patient in clinical practice.

The goal of our research is to eliminate the need for a Bucky grid through time-of-flight discrimination between primary and scattered radiation. This technique is based on the fact that the scattered radiation must propagate along a longer path in order to reach the detector. Since all of the radiation propagates at the speed of light, the scattered radiation will take longer to reach the detector than the primary radiation. A 1 cm longer path length will result in a 33 picosecond delay in the arrival of the scattered radiation. Even greater path length differences are possible, especially through multiple scattering. The first goal of our research is to determine the distribution in time of the scattered radiation which reaches the detector. We will do this through the use of a Monte, Carlo simulation. This simulation will be conducted using a code originally developed at the University of Alabama at Birmingham for designing Bucky grids. It is a modified version of EGS4 Monte Carlo code[3]. The human breast will be modeled using a homogeneous block of Lucite. Lucite has an effective atomic number of 6.5, which is in between that of the glandular tissue (7.4) and the adipose tissue (5.9) of the human breast[1]. Both the atomic electron binding effect on Compton scattering and the effect of molecular structure on coherent scattering will be considered in the simulation. From the results of the simulation, the optimum x-ray pulse length will be determined.

Time-resolved imaging has been intensively investigated in optical breast imaging [4-6]. Only a few pioneering experiments have been reported for time-resolved radiological imaging. Gordon and his coauthors first demonstrated the use of time-gated detection as a method of scatter reduction in diagnostic radiography[7]. A recent report by Gratz et. al. investigated the possible future application of time-gated imaging in volumetric CT. These experiments demonstrate that by using time-resolved discrimination it is possible to achieve significant scatter rejection, similar to that obtained with a Bucky grid but with much lower dose penalty. In these experiments a laser induced plasma was used as the, pulsed x-ray source. Two different detectors were investigated—a microchannel plate detector and an intensified streak camera. The laser produced plasma x-ray source is an excellent source of bright, short pulsed

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x radiation, well suited to an initial investigation of this technique. However, the size and cost of the equipment required to produce it are significant barriers to widespread use in this application. In addition, although laser plasma sources provide greater peak brightness than any other commonly available source, they are still not bright enough to provide single pulse imaging and the rate at which the pulse can be repeated is relatively low. The detectors used in these experiments are also better suited to demonstration than to practice. They are not practical for clinical use because of insufficient quantum efficiency for the application. The lack of either an adequate source or an adequate detector forms a significant technological bottleneck in developing a clinically feasible time-resolved mammography system. In this research we will overcome both of these difficulties through the development of a new source and a new detector. Both source and detector are modifications of those currently used in clinical practice.

2. FIELD EMISSION X-RAY TUBES

The most common source of x rays is the x-ray tube. It is relatively compact, reliable, inexpensive, and easy to use. An x-ray tube consists of an electron emitting cathode and a target anode. Both cathode and anode art contained within a vacuum envelope. The most common cathode type is a hot filament, usually tungsten. The filament is heated by passing current through it. When it becomes hot enough, some of the electrons obtain enough energy to overcome the work function of the material and escape into the surrounding free space. A large potential
difference is applied between the cathode and the anode, and the free electrons are accelerated across the gap where they strike the anode. As the electrons are decelerated by interactions with the anode material, some of the energy is converted to x rays. This process is very inefficient. Most of the energy is converted to heat in the anode. However, significant x-ray flux can also be obtained.

There are three ways in which an x-ray tube can be pulsed. First, the potential difference between the cathode and anode may be pulsed. X rays will be produced only when the potential is turned on, accelerating the electrons into the anode. Because of the high voltage required, this would be extremely difficult to do at the rates required for time-resolved scatter reduction.

The second method is to use a gate grid between the filament and the anode. Such a grid may be used to produce a relatively field-free region around the, cathode. This produces a region of space charge around the hot cathode and prevents most of the free electrons from escaping to the potential well leading to the anode. Then the gate potential maybe altered to accelerate the electrons into the anode potential well. Although this would require a much lower voltage pulse, it might still be difficult to do at the rates required. The capacitance of the gate, circuit would still be significant.

The simplest method of operating an x-ray tube in pulsed mode is simply to turn the cathode on and off. For a standard hot filament cathode this means that the rate at which the tube can be pulsed is limited by the speed with which the cathode heats and cools. While rapid, this rate is probably not fast enough for the requirements of time-resolved scatter reduction. In addition, rapid thermal cycling of the filament will stress it significantly, leading to early failure. But other electron sources are available, in particular, a field emission electron source is ideal for this application.

A field emission electron source consists of an array of sharp tips, preferably made from a low work function material, and a gate structure. When a potential difference is applied between the tip and the gate, large fields are created at the point of the tip. When these fields are large enough, electrons are pulled out of the tip material. When the geometry is correct virtually all of the emitted electrons will be captured by the potential well leading to the anode rather than going to the gate. Field emitter electron sources may, in principle, be operated at very high frequencies. The frequency is limited by the gate voltage required and the: capacitance of the gate-tip structure. By using microfabrication techniques it is possible to minimize both of these parameters. Spindt has reported the fabrication and operation of field emitters in the 10 GHz range. Even faster operation may be possible.

3. TIME RESOLVED DETECTION

For this application, a digital flat panel detector is desirable. Two categories of flat panel detectors are in use. In one, the x rays are converted to visible light through the use of a phosphor. The light signal is then collected and converted to a digital electronic signal. In the second, the x rays are converted directly to an electronic signal by using an amorphous selenium layer between two electrodes.

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There are two to three points at which the detector may be gated. First, if a means were found to provide gated transmission of x rays, the signal could be gated before reaching the detector. Second, a visible light valve could be used in the same manner in a detector which converts x rays to visible light as an intermediate step. Finally, the electronic signal output from a fast detector could be parsed by very fast electronics to eliminate the, scatter tail.

Gated transmission of x rays would provide an elegant solution to the problem. It would allow the use of any of the detection schemes used in current clinical practice, including even film. Unfortunately, the only method for gating x rays in common practice is the use of a physical shutter. In this application the shutter would have to move at a significant fraction of the speed of light, so this is not a practical solution to our problem.

Gated transmission of visible light is more plausible. Thin light valves exist, and it might be possible to operate them at sufficient speed. However, even a thin light valve might add significantly to the loss of contrast due to the fact that the visible light created in the phosphor does not necessarily emerge in the same direction as the incident x ray that produced it. In addition, most light valves work with polarized light. This requires the use of a prepolarizcr. Since the light produced in the phosphor is unpolarized, this will reduce the intensity by a factor of two. Thus, such a system would not improve, patient dose appreciably over that obtained with the use of Bucky grids. Although we have some ideas for a nonpolarized thin film light valve, they would require significant.

The remaining option is electronic gating of the detector signal. In most electronic detectors, whether they are detecting visible light from a phosphor or directly detecting the x rays, photons are detected when they are absorbed by a semiconductor material. The energy of the photon goes into the production of electron-hole pairs. The result is a small current burst. In a radiographic imaging application such as mammography, the free charges created by all of the photons collected in a pixel during, an exposure are integrated to give the intensity of the illumination at that pixel. We wish to integrate only that current which is produced during the time period at which the primary radiation from the x-ray pulse reaches the detector. To accomplish this, we will use the scheme shown in Figure 1. A control signal is generated and sent to the tube. When the tube receives this signal an x-ray pulse is generated. The same signal is sent through a delay line to the triggerable integrating amplifier. The delay line is designed to ensure that the control signal arrives at the amplifier at the same time as the x-ray pulse arrives at the detector. The control signal triggers the amplifier to begin collecting the detector current, producing a proportional voltage output. When the control signal ends, the amplifier locks in the voltage output and any additional detector current is ignored. A subsequent control signal will cause the amplifier to collect additional current from the detector, with the resulting output proportional to the integrated current.

Figure 2 shows in greater detail the way in which the triggerable integrating amplifier works. When the control signal arrives, switch 1 is closed. This allows the detector current to be collected by the charge sensitive amplifier. When the control signal ends, switch 1 opens again. The charge sensitive amplifier is cut off from the detector signal, which may be shunted to ground. The voltage signal remains at the output of the charge sensitive amplifier between control signals, and the final integrated value may be read out whenever convenient at the end of the exposure. After readout the amplifier output may be cleared by closing switch 2.

In order for this scheme to work, several conditions must be met. First, the detector must have very low capacitance. The RC time constant must be small compared with the x-ray pulse length. Fortunately, a single pixel of an imaging detector array will be very small, so the capacitance will naturally be low. The time required to open and close switch 1 must also be small compared with the x-ray pulse length. Finally, leakage current across the amplifier must be low enough that the output signal does not degrade too much during the course of an exposure. These conditions are achievable.

REFERENCES

1. Barnes, G.T. and Brezovich, I.A., "The intensity of scattered radiation in mammography", Radioloy 126, 243 247, 1978.

2. Wagner, A.J., "Contrast and grid performance in mammography", Screen-film mammography, Proceeding SEAAPM spring symposium, Edited by Bames and Prey GD, 115-134, Medical physics publishing, 1991.

3. Nelsonn, W.R., Hirayamah, Rogers DWO, "The EGS4 code system", SLAC-Report-265, Standard Linear Center, 1985.

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4. Aderson-Engle, S., Bergr, Svanberg S., Jarthman, 0., "Time resolved transillumination for medical diagnostics", Optics Letters, Vol. 15, No. 21, 1179-1181, 1990.

5. Aderson-Engle, S., Bergr, Svanberg, S,. "Effects of optical constants on time-gated transillumination of tissue and tissue-like media", J. Photochem. Photobiol. B:Biol., 16, 155-167, 1992.

6. Michielsen, K,. Raedt, H.D., Garcia, N,. "Computer simulation of time-gated transillumination and reflection of biological tissues and tissue-like phantoms", Med. Phys., 42 (11), 1688-1695, 1997.

7. Gordon,C.L., Yin, G.Y., Lemoff, B.E., Bell , P.M., Barty, P.J., "Time-gated imaging with an ultrashort-pulse, laser-produced-plasma x-ray source", Optics Letters, Vol. 20, No. 9, 1056-1058, 1995

8. Spindt, C.A., Holland, C.E., Schwoebel, P.R., and Brodie, I,. "Field-Emitter-Airay Development for Microwave Applications", Journal of Vacuum Science and Technology - B, Vol. 14, No. 3, pp 1986-89, May/June 1996.

9. Spindt, C.A., Holland, C.E/, Schwoebel, P.R., and Brodie, I.,"Field-Emitter Array Development for Microwave Applications (H)", Journal of Vacuum Science and Technolo - B, Vol. 16, No. 2, pp 758-761, Mar/Ap

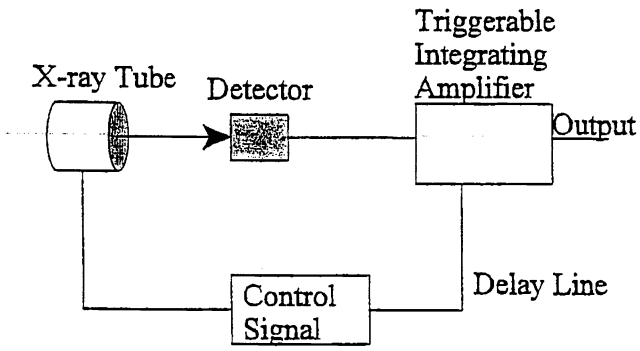


Figure 1. Diagram of electronic discrimination scheme.

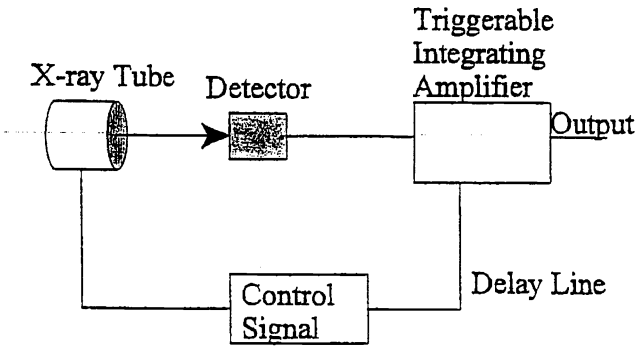


Figure 2. Detail of the triggerable integrating amplifier.

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