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Design of a Mammary gland Specific Gamma Imaging System (MSGIS)

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Abstract-

In conventional X-ray mammography, the presences of a small lesion/tumor at the early stage of cancer are not definitely detected because of the complex anatomy of the breast tissues. On the other hand, the low resolution image produced by conventional scinti-mammography does not differentiate between very small benign and malignant tissues. Successful development of Mammary gland Specific Gamma Imaging System (MSGIS), a kind of gamma camera system, may lead to the early detection of breast cancer. A low cost such a camera system is being developed using electron multiplying CCD (EMCCD) as the detector.

I. Introduction

Mammary gland Specific Gamma Imaging System (MSGIS), a kind of gamma camera system, is an effective diagnostic tool especially for patients, who have dense tisues, suspicious areas on a mammogram, lumps that can be felt but not seen with mammography or ultrasound or having implants or scaring from previous surgeries. Ultrasound or MRI can offer additional diagnostic information that is sensitive to changes, whereas MSGIS is specific to determine metabolic changes within the breast that indicates suspicious cells. The MSGIS camera allows imaging with no compression of the tissues. The advantages of scintimammography are: (i) improves evaluation of positive mammograms, (ii) differentiates between benign and malignant tissue, (iii) detects small tumors in cases where mammograms are difficult or impossible to read, (iv)could reduce the need for biopsy and (v) allows images where none were possible before. Unlike standard devices, such as X-ray mammography, this imaging detector is capable of capturing close views of the tumor, and increasing accuracy in detection and localization of small lesions. This paper describes a cost-effective new camera system.

II. Design and Methodology

MSGIS adopts molecular breast imaging technique which depends on the localization of radiotracer. An example of radiotracer is technetium-99m (Tc-99m), which by the way is the most common radioisotope used for diagnostic purposes today in medicine. Tc-99m is most widely used because it has half-life of 6 hours which allow sufficient measurement and receive less radiation by body organs. The two dimensional pixilated camera captures these gamma rays from different position of the target and could form a high resolution 2-D image based on the spatial intensity distribution of the gamma ray. The basic design of the camera consists of three main sections depicted in the Fig-1. The first section is the collimator, can be thought as a focusing device. Gamma-ray photon energy to electrical signal conversion could be achieved by the two stage conversion like scintillator in 1st stage and PMT/Diode/CCD in the second stage. The energy resolution and rise time of the 1st stage detector, the signal to noise ratio (SNR) and speed of the 2nd stage detector are the prime consideration for the design of the gamma camera. The radio medicine, used for the purpose, emits the gamma-ray of this energy 140KeV. Our desired objective is to extract the140KeV gamma-ray information from the electrical pulse output of 2nd stage.

TABLE I. Specification of photoelectric converters

Converter Type	APD Array	PSPMT
Spectral response	340-720nm	300-650nm
Peak Sensitivity Wavelength	550nm	420nm
Quantum efficiency	70%	24%
Dark Current	10nA	0.05nA
Terminal Capacitance	10pF (approx.)	
Gain	50 or more	10^5 at -700V
Element Size	2.3mm	2.8x 2.8 mm2
Crosstalk		5%

A small tracing element is injected which is absorbed by all the cell of the body for gamma imaging. Because cancerous cell have higher rate of metabolic activity, the tracing usually concentrates in these cells, if present. The tracing agent emits gamma rays which are detected by the imaging system.

The present technology of gamma-ray detection is based on combination of scintillation crystal and photoelectric converters. Typically CsI(Tl) crystals with photomultiplier tubes are used [1]. Following the design specification for a high-resolution gamma-camera with high counting efficiency

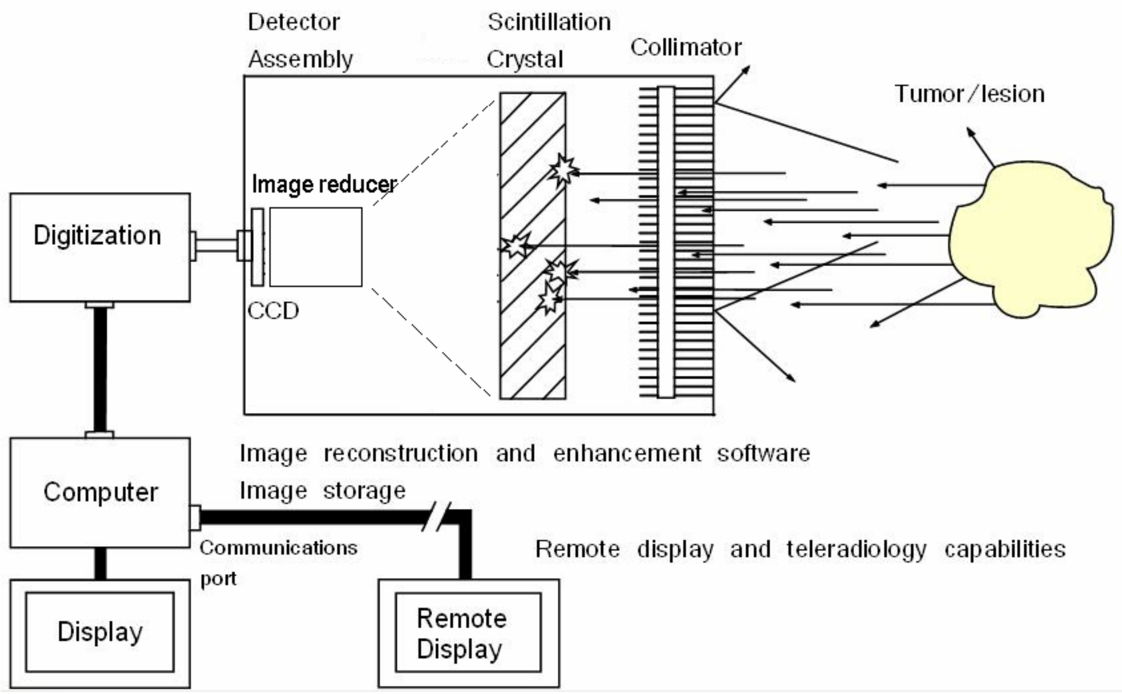

Fig. 1 Schematic of a Gamma camera

we decided for a combination of Hamamatsu make flat panel multimode photomultiplier tube assembly and columnar CsI(Tl) scintillator matrix.

The crystal needs to be suitably coupled to a Hamamatsu H9500 PMT with specifications given in table I. The working voltage is to be suitably chosen to obtain gain of ~10^6. The suitable read system electronics board with multi-channel amplifier, array of data storage, multiplexed output and control logic needs to be developed. The output of the board needs to subsequently digitized and transferred to a computer.

III. Detector to be used

The magnetic field interference and high voltage bias has obviated the use of PMT as the 2nd stage device. Hence, diode detector or CCD (charge couple device) has to be used in place of PMT. Since the gain of PIN diode is unity, detecting a photo peak, resembles the statistical distribution of the scintillation corresponds to a genuine 140 KeV gamma ray, is almost impossible. The background noise dominates the signal and reduces the quality of the image. The use of PMT is the limiting factor to obtain higher spatial resolution and robustness as they are bulky, fragile and susceptible to magnetic field. Therefore we would also investigate an improved gamma tomography detector based on semiconductor avalanche photodiode (APD) in combination with the fast and high optical gain scintillator crystals (LYSO) having specification given in table II [2]. The analog signal processing circuit suitable for this combination needs to be developed. The charge generated in APD during the photoelectric conversion process needs to be converted to a voltage pulse in the preamplifier stage. In typical configuration preamplifier is followed by a CR-CR pulse shaping amplifier and subsequently followed by a gain adjustable amplifier. The preamplifier design is the most critical since it determines electronic noise and thus energy resolution.

TABLE II. Specification of scintillation crystal

Scintillation material	Thallium doped Cesium Iodide	Lutetium Yttrium Orthosilicate
Spectral response	340-720nm	350-600
Peak Wavelength	550nm	420nm
Number of pixel	16x16	4x8
Pixel dimension (mm^3)	2.7 x 2.7 x 10	1.6 x 1.6 x 5
Pixel size	3.0mm	2.3 mm
Separation material	Teflon	Teflon
Wall separate size	0.23 mm	0.7 mm
Light yield (photon/keV)	54	32
Decay time	1000	41
Way to stop photon of 662keV	2 cm	1cm
Hygroscopic	Slightly	No

There are a few commercially available monolithic amplifiers IC that are suitable for scintillation pulse amplifier. For high spatial compactness and less power consumption use of ASIC (amplification specific integrated circuit) may be the most suitable. In view of the discussions above, it appears that CCD may be a best choice when we are targeting for a very high resolution image. The required electrical/optical specification for CCD, considering the CSI (TI) scintillator detector to be coupled with it is as follows: (i) 350-850 nm Spectral Response with peak sensitivity at 550 nm, (ii) High Quantum Efficiency, (iii) low terminal capacitance and (iv) element size in order of mm or less.

Presently sub-milimetric image resolution using gamma camera is achievable by employing the principle of pinhole magnification. Compact high-resolution camera using solid state based non-intensified electron multiplying charge couple device (EM-CCD) in combination with CsI(Tl) micro-scintillators matrix. Such a detector effectively reduces readout noise to less than one electron RMS. The back-illuminated EMCCD has single photon detection capability without an image intensifier combined with greater than 90% QE (quantum efficiency). It enables charge to be multiplied on the sensor before it is read out with 512 x 512 frame transfer facility. The absolute EMCCD gain of the camera can be varied linearly from unity up to a thousand times directly through the software. The presently available system offers up to 10 MHz pixel readout rate, conventional amplifier outputs with minimized dark current with thermoelectric cooling. In such CCD-based detectors individual scintillating events can be detected in photon counting mode [3]. The dark current problem of EMCCD may be reduced by cooling the array's improving the sensitivity of the CCD to low light intensities even for ultraviolet and visible wavelength (for details see [4].

IV. Summary

The objective preliminary is to develop a gamma imaging system based on three different methodologies i.e. combination of scintillator and detectors and to study and compare the performances like spatial, energy resolution, uniformity and count rate performances of the system

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