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Scintillating fiber optics and their application in radiographic systems

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The Ohio State University, 1992

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SCINTILLATING FIBER OPTICS AND THEIR APPLICATION IN RADIOGRAPHIC SYSTEMS

DISSERTATION

Presented in Partial Fulfillment of the Requirements for the Degree
Doctor of Philosophy in the Graduate School of The Ohio State
University

By

Hua Shao, B.S. M.S.

* * * * *

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1992

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A handwritten signature in dark ink, appearing to read "Don W. Miller", is written over a horizontal line.

Advisor

Nuclear Engineering Program

Dedicated to
Leo Shao,
my son

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& Nuc. Instru. & Methods, A2999 (1990) 528-533

H. Shao et al., "The Application of Scintillating Fiber Optics in the Scantech DADR System", 1990 IEEE, Nuclear Science Symposium, Vol. 2, 1075-1084

D. Miller, and H. Shao, "Scintillating Fiber Optical Radiation Detectors", ANS Transactions, Vol. 62, Nov. 1990, 204

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LIST OF SYMBOLS

$*$	convolution function
$**$	2-D convolution operation
	attenuation constants of waves in the direction of x and y
α	the light attenuation coefficient for the fiber core in the z direction
β	the reflection coefficient of the core/cladding interface
B_e	electron binding energy
CCD	charge coupled devices
$\text{comb}(x/X)$	comb function separated by $1/X$ due to the finite sampling
χ	pair production linear attenuation cross section
D	system operator or blurring function, or the diameter of the fiber
DADR	Diode Array Digital Radiography
DE	detector efficiency
DVM	digital voltage meter
	element of area in the fiber core cross section
	element of length along the fiber, and
	element of solid angle,
	element of the wavelength
δ	delt function

E	intensity of field, or photon of energy
E'	photon energy after the interaction
EMA	extra mural absorber
EQSPEC	equivalent spectrum
ERF	edge response function
ϵ	light absorption coefficient of the EMA
F	Fourier transform operator, or the total flux of light
f	the upper frequency in Hz at which the response is 70.7% of the middle band response, or fraction of the incident quanta which are attenuated
F^{-1}	inverse Fourier transform operator
FFT	Fast Fourier Transform
FOM	the figure of merits
FWHM	full width at half maximum
$g(\bar{r})$	image function
γ	the angle between the y-z plane projection of the ray and z-axis
η_a	photon energy absorption efficiency
η_c	light conversion efficiency
η_t	light transmission efficiency
$G(u,v)$	two dimensional Fourier transform
g_j	the gain of jth stage of the system
$g_s(x,y)$	sampled image function
$H(u,v)$	spatial transfer function of the system in the frequency domain

$H_n\{.\}$	Hankel transform of order n
$\bar{H}(\rho)$	normalized frequency response
h_0	point spread function at center fiber
$h_i(r, d)$	point spread function at ith fiber due to the scintillating at point d
$h(\bar{r}, \bar{\xi})$	impulse response function or point spread function
$h(\bar{r}, \bar{\xi})$	impulse response function or point spread function
$\bar{h}$	counts per unit in background
$\Delta h / \bar{h}$	threshold contrast
I_0	intensity of incident photons
I	input function
$I(t)$	the intensity of I_0 after passing a thickness t
$I_{trans.}$	total light transmitted by a straight fiber waveguide
ITS	the integrated tiger series
$I(\theta, \phi, z, \lambda)$	distribution of the scintillating light in the medium of the fiber core
$J_0(2\pi r \rho)$	Kernel of the Fourier-Bessel transform for the functions having circular symmetry
ϕ	the angle between the projection of propagation onto x-y plane and x-axis
k_0	free space wave number
κ	wave propagation direction expressed in terms of Kappas
Λ	total attenuation coefficient which accounts for reflection and absorption losses
LSF	line spread function

$\mu(E)$	the photon linear attenuation coefficient at energy E
MTF	modulation transfer function
N	the number of photons per pixel element impinging on the detector, or photon interaction rate
n	the number of reflections in unit length of the fiber
NA	numerical aperture
NDT	Non-Destructive Testing
N_m	number of modes
N_x, N_y	number of modes in x and y direction respectively
n_0	refractive index of the medium outside the left face
$n_{\text{core}}, n_{\text{clad}}$	refractive indexes of the core and cladding respectively
$p(z)$	interacting density function of gamma-ray photons at the point z
PET	positron emission tomography
PMT	photomultiplier tube
PSF	point spread function of the system
θ	transmission angle of scintillating light ray, or the angle between the direction of propagation and z-axis
θ_c	the critical angle for total reflection
θ_e	light exit or the acceptance angle
θ_p	light propagation angle
θ_m	maximum angle of acceptance
θ_s	photon scattering angle measured from its initial direction
r	distance

$r(\text{mg} \cdot \text{cm}^{-2})$	range of electrons
$\hat{r}$	ray propagation direction
S	intensity of gamma source
σ	Compton scattering linear attenuation cross section
SNR	signal-to-noise ratio
SPECT	single photon emission computer tomography
S(E)	x-ray energy spectrum
$s(\bar{r})$	source distribution function
T	kinetic energy of secondary electrons
T_{cp}	kinetic energy of Compton scattering electron
T_{pp}	kinetic energy of electron-positron pair
T_{p}	kinetic energy of photoelectron
t	the transmission function which depends on the properties of the fiber and the orientation of the ray
τ	time constant of luminescence, or photoelectric linear attenuation cross section
$t_{\perp}, t_{\text{II}}$	transmissivity of the light, the paralleled component denoted by subscript II and the perpendicular component by $\perp$
u, v	spatial frequencies
Ψ_{c}	maximum, or the critical angle

CHAPTER I

INTRODUCTION

1.1 Objective

The overall goal of this research is evaluation of scintillating fiber optics (SFO) and their applications in the imaging of x-rays. Two sub-objectives supportive of this objective are development of an analytic model to predict the performance of SFO detectors and completion of an experimental program to measure the physical parameters that determine the performance of SFO radiation detectors when used to image spatial distributions of x-rays. A corollary objective is a computer evaluation of SFO in high energy (>500 keV) x-ray imaging.

1.2 Methodology

In this research, the characteristics of SFO detectors were investigated using a combination of theoretical and experimental

The general theory relevant to x-ray imaging and scintillating fiber optical detectors was first developed.

An experimental evaluation of the inherent properties of SFO as radiation imaging detectors was then evaluated. In this study, two types of Tb_2O_3 based SFO were studied using monoenergetic gamma rays, and both low and high energy x-ray sources.

The study evaluated the overall detector efficiency by combining experimental measurements and analytical Monte Carlo methods. It was found that the detector efficiency was low when compared to phosphor screens. This, however, was due to low light production and not radiation absorption efficiency. Therefore, by using light amplification, low detector efficiency was compensated for, without effecting spatial resolution.

The inherent spatial resolution of SFO was evaluated experimentally at both low (hundreds of keV) and high (a few MeV) photon energies by using edge response functions. At low energy levels, the spatial resolution is 8-9 lp/mm. At high energy levels, although uncertainties due to backscattering and the experimental configuration were introduced, spatial resolution for a 10 mm SDD imaging is estimated to be 4-5 lp/mm including the degradation from strong backscattering.

The SFO response time constant was investigated with a high energy x-ray pulse (30 ns) and determined to be on the order of milliseconds. The scintillating light spectrum, optical coupling and linearity were also studied.

Research to evaluate the use of scintillating fiber optics (SFO) to replace traditional phosphor screens as an imaging detector in a digital radiographic system is presented. A SFO-based detector was constructed for a Diode Array Digital Radiography (DADR) system [1]. System image quality was evaluated with five characterization criteria: spatial resolution, signal-to-noise ratio (SNR), large-area contrast detectability, responsivity, and inspection task performance. It has been concluded from this evaluation that the spatial resolution, SNR and large-area contrast detectability are superior to that observed with phosphor screens. However, responsivity, with SFO is much lower than measured with phosphor screens. In the energy range of hundreds keV x-rays, although the x-ray absorption efficiency of SFO is high, the overall efficiency is lower than for phosphor screens. By adjusting the gains of an image intensifier and an electronic amplifier, an adequately large signal output for a feasible x-ray dose input was obtained with the SFO system. The SFO system has produced images of electronic assemblies and met the requirements of the ASTM 801E #4. The experimental results indicate that the SFO detector is comparable to phosphor screens at low x-ray energies ($< 300 \text{ kV}_p$), and is expected to have superior performance for higher energy photon detection.

Based on the physical analysis and experimental results, a computer simulation was developed and used to investigate the efficiency and spatial resolution of the SFO. The simulation is a combination of a Monte Carlo method and an analytic model. The Monte Carlo method calculated the energy deposition in the SFO detectors and the analytic model simulated

the light transmission properties of the scintillating fiber optics. By changing the input parameters, the computer simulation can calculate the detector efficiency and the spatial resolution of SFO detectors at any photon energy level. The computer simulation results were compared with the experimental results, such as the effects of the photon energy and detector thickness on the spatial resolution. The computer model was also developed to assist in parametric analysis of the interrelated effects of several inherent and technological limitations. Therefore, it is valuable for design purposes for SFO detectors.

A quantitative method-figure of merit (FOM) was developed to determine the imaging capability of the SFO detectors. This method combines the spatial resolution and the noise structure information to provide a measure of the detectability of small defects on a uniform background. The simulation indicates that the SFO detectors have an overall better performance than observed in phosphor screens, and will have a much better performance when used for higher energy photon detection.

1.3 Motivation

Current x-ray image detectors used in radiographic imaging systems are poor detectors. Enhancing their detection efficiency requires thicker detectors which inevitably results in degradation of spatial resolution from lateral light dispersion. Scintillating fiber optics combine scintillating material with a dielectric waveguide. This configuration

permits an increase in thickness of the scintillating material without significant light dispersion , thereby preserving spatial resolution with increased detection efficiency. This characteristic is advantageous for imaging of high energy x-rays in which a thicker detector is required.

Many kinds of x-ray image detectors are widely used in non-destructive testing, medical imaging, and experimental physics. A brief review of different x-ray imaging detectors, their advantages and disadvantages is presented as follows.

X-ray image detectors can be divided into two classes [2].

(a) Time deterred detectors

The detectors use physical or chemical changes in a recording medium. These are not real time imaging detectors and therefore are not used in real time systems.

(a.1) Silver halide photographic plates

They provide high output and good spatial resolution. However, they can only count a limited number of photons before saturation, and therefore the image statistics cannot be improved.

(a.2) Photoresistors

They have a resolution down to 10 nm and large fields, but they are used only for soft x-ray energies and their sensitivity is very low.

(b) Real time detectors

The detection of x-ray is based on the detection of secondary particles. These are real time detectors.

(b.1) Multiwire chambers

1-D or 2-D multiwire chambers have high efficiency and output, but their spatial resolution and counting rate is limited.

(b.2) Photodiode arrays

Photodiode arrays can be used in direct x-ray exposure in the 1-10 keV range. They show good efficiency and spatial resolution. However, they are easily damaged by the x-rays and use limited to low energy applications only.

(b.3) Channel plates

The microchannel plates can be used directly as an x-ray to electron converter. However, this device must be held under a high vacuum and their quantum efficiency is very low.

(b.4) Charge coupled device arrays

X-ray sensitive charge coupled devices (CCD) are photosensitive arrays made on silicon wafers by large scale integrated circuit technology. The photosites collect electrons released by incident radiation. Charges that reach the depletion region are kept out of the device directly into amplification and recording electronics. CCD x-ray image detectors have been widely used in the keV x-ray energy range.

(b.5) Phosphor screens

Phosphor screens based on the visible fluorescence of a scintillating powder are the conventional x-ray converter used in radiographic systems. They show a high quantum efficiency when a thick layer is used. However, they are poor detectors for high energy photons. This is a consequence of the low interaction rate of high energy photons within the detector. To increase the interaction rate, the detector must be made

thicker. Unfortunately, light dispersion in thick detectors leads to a degradation of spatial resolution due to scattering of scintillating light as shown in Figure 1.1 (a).

Comparing these different devices, it is found that no ideal x-ray detector currently exists, although some good solutions have been proposed in the soft x-ray energy range (< 1 keV) [2]. It has also been indicated that the critical device limiting the performance of real time radiographic systems is the x-ray to light image converter [3]. Therefore, to find an appropriate x-ray detector (for high energy x-ray) is of significant value.

Figure 1.1 Light Dispersion in Phosphor Screens and Scintillating Fiber Optics

The ideal image detector for harder x-rays should have the following properties:

- (1) it can be made as thick as necessary to maintain the detection efficiency and signal-to-noise ratio;
- (2) spatial resolution should not be a function of detector thickness.

Scintillating fiber optics (SFO) can meet these requirements by including the scintillation material within the fiber core. Some of the light from the scintillation is trapped in the fiber where it is generated - preserving the spatial reference of the event by minimizing light dispersion as shown in Figure 1.1(b). Thus, the scintillating fibers play a dual role: first as a scintillator and second as a light pipe. The advantage of the SFO is that it can be made thick to increase the interaction rate while retaining high spatial resolution by channeling the generated light. Scintillating fiber optics with the individual fiber diameter as small as 10 μm and thickness up to several centimeters have been made. These facilitate high spatial resolution and a high interaction rate [4,5].

A SFO is usually configured in an arrays and then coupled into an array of photodiodes for 2-D imaging of radiation. Initial experiments have demonstrated the potential for using SFO in real time x-ray radiography [6]. The results of this research indicates that the SFO detectors will offer significant improvement for high energy photon imaging, and superior to traditional phosphor screens.

1.4 State of the Art Scintillating Fiber Optics

As early as the 1950's, L. Reiffel and N. S. Kapany from the Armour Research Foundation [7], as well as independent work by other groups [8,9] used the light piping techniques similar to those used in fiber optics to collect light emitted from the track of charged particles interacting with a phosphor. Simple extension of the ideas of Conversi [10], and Barsanti et al.[11] made it obvious that chambers consisting of an array of scintillator filaments serving as light pipes could also give three-dimensional information on particle tracks.

In a 1981 publication, S. R. Borenstein, et al, reported their effort to develop scintillating fibers and construct scintillating fiber based hodoscopes [12,13].

In 1985, E. Bigler and F. Folack [2] reported their work on liquid scintillating fiber arrays for high resolution x-ray imaging. Their detector was an array of optical fibers, the core of which had been replaced by a high index fluorescent material.

In 1987, Michihiro Ito, et al., reported their new scintillating detector developed by growing CsI(Na) crystals on a special fiber plate [14]. The individual crystal had a thickness up to 0.12 mm, and the diameter of 0.02 mm. The spatial resolution measured by electron beam excitation was approximately of 0.03 mm FWHM.

A new configuration was proposed for the realization of an x-ray detector by M. Bourdinaud and M. Laguesse[15]. It is based on an

inorganic scintillator optically coupled to a ribbon of fluorescent optical fibers in such a way that a high detection efficiency may be obtained. X-rays coming from the objective to be analysed reach the scintillator in a direction parallel to the fiber axis. A high x-ray conversion efficiency was achieved due to increasing scintillator thickness. The absorption of an x-ray produces excitation in the scintillator, and a fraction of the resulting visible photons reach the fluorescent fiber ribbon. In turn, these photons excite the fluorescence of the fibers. This principle allows one to realize a linear x-ray detector. However, a scanning device is needed in order to reconstruct a two dimensional image.

Scintillating fibers have been used widely as particle track detectors, as reported by Dr. Randy Ruchti [4,5,16,17,18]. In Ruchti's apparatus, high energy particles are incident on a fiber optic plate. As they travel through the fiber plate perpendicular to the fiber axes they deposit energy in the individual fibers and emit visible light. A light image detector at the end of the fiber optic plate records which fibers emitted light and store the resulting image in a computer.

For several years, The fiber tracking group (FTG) has been developing many types of scintillating fiber optics for tracking applications [19]. At CERN, a large scintillating fiber detector consisting of approximately 60000 scintillating plastic fibers of 1 mm diameter and 2.4 mm length is used as a tracking and preshower detector. The scintillating fiber detector has two separate functions: the inside layers give a precise localization of the particles and the outside layers initiate the electromagnetic showers by conversion in a lead sheet [20]. Over the

last few years, scintillating fiber detectors have been used extensively in high energy physics experiments [21,22,23,24].

In medical imaging, high spatial and time resolution gamma ray detectors have been developed using plastic scintillating fibers coupled to position sensitive vacuum photomultipliers [25,26]. These detectors can significantly improve the spatial resolution, time resolution, and efficiency of both SPECT and PET, extend the application of these technologies into new fields of medical research, and improve currently existing methods of medical diagnosis. Initial results of testing these detectors showed a better than 1 mm position accuracy and better than 10 nsec time resolution.

In industrial radiography, R. Polichar reported the use of scintillating fiber optics in high resolution real time radiography [3]. The study covered the x-ray potentials between 20 kV and 320 kV. Illustrations are provided on application of this new technology to inspection of aerospace, electronic and ordinary samples. The varian ER real time system uses rare earth type scintillating fiber optic screens when imaging x-ray from 1 MeV up to 15 MeV [27]. The exclusive screen is composed of an array of millions of 25 micros scintillating fibers, producing high energy x-ray conversion while preserving excellent spatial resolution.

Overall, scintillating fibers of many different compositions have been manufactured. Scintillating materials can be solid or liquid, organic or inorganic. The scintillating fiber optics has found many applications; including Calorimetry, tracking detectors, spectrometers, hodoscopes,

neutron detectors and image detectors. Scintillating fiber optics have been used in biological, medical, industrial and high energy physics applications.

At The Ohio State University, we have been collaborating with the Scantech Corporation to evaluate SFO as radiation image detectors. As mentioned before, conventional image detectors lose spatial resolution when designed for high energy photon detection. On the other hand, scintillating fiber optics have an intrinsic nature that can maintain the spatial resolution even with high energy photons. The overall goal of our work is to evaluate the feasibility of substituting SFO for phosphor screens in radiographic x-ray systems. Two types of Terbium Oxide (Tb_2O_3) based inorganic scintillating fiber optics have been tested in our research. These scintillating fiber optics have a greater mass/unit area than conventional phosphor screen x-ray image detectors while retaining similar spatial resolution.

CHAPTER II

GENERAL THEORY

2.1 Introduction

The application of scintillating fiber optics to the imaging of high energy photon spatial distribution requires a basic understanding of detector physics and system analysis method. The purpose of this chapter is to provide readers the fundamental introduction of the physics and methods used and how they are integrated to analyze and apply SFO as an imaging modality.

The chapter begins with a discussion of scintillating fiber optics, introducing an assessment of the materials used in SFO, the transmission of light and loss mechanism. Next, the topic of linear system and two dimensional spatial imaging is addressed. The discussion includes the application of Fourier transform analysis to digital spatial imaging and its relation to the performance of SFO.

The physics of radiation interaction with matter are reviewed with emphasis on the interaction mechanism that affects spatial distribution. Then a discussion introduces the integrated tiger series (ITS) Monte Carlo

code for modeling of high energy photon and electron transport and the application of this code to modeling the physical processes taking place in SFO.

Finally, a summary of the theoretical description of characteristics of the scintillating fiber optical detector is presented. The six factors that determine the performance of SFO detectors are introduced and related to the measurement of high energy photons. A mathematical expression describing the light transmission properties of SFO is developed using both modal and ray approaches.

2.2 Introduction to Fiber Optics

Fiber optics have many desirable characteristics and thus have been used widely in communication, instrumentation and control. Among these properties are: high bandwidth, immunity from electromagnetic interference, small physical size, flexibility and radiation resistance.

Vitreous SiO_2 glass composes modern fiber optics. V- SiO_2 is an amorphous material formed by cooling high energy SiO_2 fast enough to preclude crystal growth. The defining characteristic of an amorphous material is that its structural arrangement of atoms exhibits no repeated unit cells. The fundamental building block of v- SiO_2 , however, remains essentially perfect $\text{Si}(\text{O}_{1/2})_4$ tetrahedra [29]. The relative orientation of the tetrahedra is characterized by a dihedral angle and a statistical distribution of Si-O-Si angles. Although locally the atomic structure has

considerable order due to the directed covalent bonds and invariant bond lengths, this construction is referred to as a continuous random network.

Optical fibers have been undergoing a revolution in purity of materials and manufacturing techniques. Silica, pure or doped, composes the core and cladding of majority of optical fibers [30]. Common dopants are GeO_2 and P_2O_5 which raise silica's index of refraction, and F and B_2O_3 which lower it [31].

Information is transmitted along fiber optics in the form of light. In general, a optical fiber cross section consists of a central dielectric material known as the core surrounded by a cladding dielectric of slightly lower refractive index. Fiber optics are divided into two classes.

Singlemode and multimode. The division is based upon the number of Hermite-Gaussian modes that will propagate down the fiber. Multimode fibers are further subdivided into step and graded index fibers. In a step index fiber both the core and the cladding have distinct refractive indices, whereas, graded index fibers have refractive index profiles highest at the centerline and decreasing, usually parabolically, outward [32].

Light propagation down a fiber is analyzed by either electromagnetic field theory or by ray optics. Due to the extremely small core size singlemode fibers are usually described by electromagnetic field theory [33]. Multimode fibers, on the other hand, can be described by traditional ray optics. In step index fibers light rays propagating through the fiber core will experience complete internal reflection on the core/cladding interface and be transmitted down the fiber provided they intersect the interface at less than the critical angle. In graded index fibers

the light rays are continuously bent towards the fiber axis by the refractive index profile, and therefore transmit any light rays for which the maximum deviation from the fiber centerline is less than the fiber radius.

Data transmission rates in fiber optics are limited by dispersion effects. The none parallel rays will have longer pathlengths and therefore be delayed relative to the parallel rays. This causes pulse spreading and is a fundamental limit of multimode fibers. However, multimode fibers can carry more light power and have less problem in optical coupling than singlemode fiber. For this reason, waveguides are always made by multimode fibers.

The two loss mechanisms for light photons traveling through optical fibers are scattering and absorption. The two scattering mechanisms are Raleigh scattering and Raman scattering. Raleigh scattering is the result of scattering due to the fluctuations in the atomic density of the transmission media, and decreases with increasing wavelength as λ^{-4} [34]. Raman scattering involves the emission or absorption of a phonon and as such shifts the incident photon frequency by an amount dependent on the vibrational modes of the media [35]. Optical absorption consists of a photon exciting an electron in a material from one energy level to another. Thus in order to be transparent no electron energy states can exist in the material within the energy of an incident photon above the highest ground state electron energy in the material.

In crystals the allowed electron energy levels are represented by a band structure. Energy gaps in the electron energy band structure arise

because reflected electron waves from one atom interfere destructively with the electron waves reflected from a nearest neighbor atom for electron within the gap [36]. Since short range order determines the electronic density of states, the band model applies to amorphous materials, and energy gaps within spectrum of electron energy states are expected. However, the randomness of atomic structure over ranges larger than the SiO₂ tetrahedra prevents a conventional band gap from existing, and electron wavefunction in the unfilled higher energy states may be localized [37].

2.3 Linear System and Image Theory

2.3.1 Definition of Linearity

Many of the phenomena found in imaging systems exhibit linear behavior. For example, if double the intensity of the emitting x-ray source of the radiographic system, the resultant image intensity will double. Also, if an image is recorded due to a first source and then another due to a second source, the image due to both sources acting simultaneously is the sum of the individual image. These are the two well known properties, scaling and superposition which define a linear system. The formal expression is

$$D[aI_1(\bar{r}) + bI_2(\bar{r})] = aD[I_1(\bar{r})] + b[D I_2(\bar{r})], \quad (2.1)$$

in which D is the system operator, a and b are constants, and I_1 and I_2 are the two input functions [38].

Typically, an imaging system is described mathematically as the system operator D on a source distribution $s(\bar{r})$ which produces an image

$$g(\bar{r}) = D[s(\bar{r})], \quad (2.2)$$

where, the system operator D is a blurring function that smears the original image. As will be discussed subsequently, D can be a convolution operation with the point spread function (PSF) of the system. Thus, Equation 2.1 claims that the weighted sum of two blurred images is equal to the weighted sum of images that are blurred by the system function. This powerful concept allows us to decompose the image, operate on the individual parts of the system function, and then sum to obtain the desired output image.

It must be realized that in Equation 2.1, D is a linear system operator, and therefore can only apply to a linear system, or a nonlinear system that can be approximated by a linear form. Equation 2.1 does not apply to a nonlinear system, such as photographic film that exhibits a saturation value beyond which a further increase in input intensity results in no change in the recorded intensity.

2.3.2 δ Function and Point Response Function

By using the system's linearity property, the response of any linear system can be expressed in a manner which provides significant physical

insight. The input function of a system is first decomposed into elementary functions. The response to each elementary function is found and summed to obtain the output function. The most convenient elementary function for the composition is the delta function δ . The formal definition for the delta function is [39]

$$\int_{-\infty}^{+\infty} \delta(\bar{r}) d\bar{r} \equiv 1, \quad (2.3)$$

and

$$\delta(\bar{r}) d\bar{r} = 0 \quad \text{for} \quad \bar{r} \neq 0. \quad (2.4)$$

For example, the source distribution can be assumed to consist of a collection of point sources, as given by

$$s(\bar{r}) = \int_{-\infty}^{+\infty} s(\bar{\xi}) \delta(\bar{r} - \bar{\xi}) d\bar{\xi}, \quad (2.5)$$

where, the delta function at $\bar{r} = \bar{\xi}$ has shifted out the particular value of s at that point. The output function is given by

$$g(\bar{r}) = D[s(\bar{r})] = D\left[\int_{-\infty}^{+\infty} s(\bar{\xi}) \delta(\bar{r} - \bar{\xi}) d\bar{\xi}\right]. \quad (2.6)$$

Since linearity is assumed, $s(\bar{\xi})$ can be structured as a weighting factor on each delta function so that

$$g(\bar{r}) = D[s(\bar{r})] = \int_{-\infty}^{+\infty} s(\bar{\xi}) D[\delta(\bar{r} - \bar{\xi})] d\bar{\xi}. \quad (2.7)$$

Within the integral the system response is a delta function. This is known as the impulse response function or point spread function, and given by

$$h(\bar{r}, \bar{\xi}) = \text{PSF}(\bar{r}, \bar{\xi}) = D[\delta(\bar{r} - \bar{\xi})]. \quad (2.8)$$

Therefore, a superposition integral is obtained based on the fundamental concept of linearity property,

$$g(\bar{r}) = D[s(\bar{r})] = \int_{-\infty}^{+\infty} s(\bar{\xi}) h(\bar{r}, \bar{\xi}) d\bar{\xi}. \quad (2.9)$$

Equation 2.9 characterizes an output function $g(\bar{r})$ completely in terms of its response to impulse. Once $h(\bar{r}, \bar{\xi})$ for all input coordinates is known, the output due to any input can be obtained.

2.3.3 Space Invariance

A simplification can be added to the point spread function if the system is spatial invariant. The impulse response of the spatial invariant system merely shifts its position for different input points, but does not change its functional behavior. In this case, the PSF is dependent solely on the difference between the coordinates and the position of the impulse as given by

$$h(\bar{r}, \bar{\xi}) = h(\bar{r} - \bar{\xi}). \quad (2.10)$$

The output function becomes

$$g(\bar{r}) = \int_{-\infty}^{+\infty} s(\bar{\xi})h(\bar{r} - \bar{\xi})d\bar{\xi}. \quad (2.11)$$

For a two dimensional (2-D) geometry,

$$g(x, y) = \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} s(\xi, \eta)h(x - \xi, y - \eta)d\xi d\eta. \quad (2.12)$$

This is the convolution function,

$$g = s * h. \quad (2.13)$$

The convolutional relationship clearly indicates the blurring of the s input function by the impulse response h , and it states that the point response function

$$h = h(x, y), \quad (2.14)$$

is a unique imaging characteristic of the imaging system, where x and y are coordinates in the image plane.

In principle, $h(x, y)$ might be measured at various points by means of a small aperture and photosensitive device. The aperture should be very small (to approximate a point) to get high accuracy. However, the smaller the aperture, the smaller the energy entering the aperture and the greater the difficulty for an accurate measurement.

2.3.4 Line Spread Function and Edge Response Function

The experimental measurement is more feasible if the receiving plane is scanned with a long, narrow slit, or scanned through a sharp edge. The line spread function (LSF) is obtained by measuring the energy received through a slit. Similarly, the edge response function (ERF) can be found by using an edge. The relation between LSF and ERF are

$$\text{ERF}(X) = \int_{-\infty}^{+\infty} \text{LSF}(x) dx. \quad (2.15)$$

LSF and ERF are both one dimensional (1-D) functions. For example, if the distributions which vary only as a function of x (the y coordinate is assumed fixed at y_0), Equation 2.12 becomes

$$\begin{aligned} g(x, y_0) &= \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} s(\xi, y_0) h(x - \xi, y_0 - \eta) d\xi d\eta \\ &= \int_{-\infty}^{+\infty} s(\xi, y_0) \left[\int_{-\infty}^{+\infty} h(x - \xi, y_0 - \eta) d\eta \right] d\xi. \end{aligned} \quad (2.16)$$

The line spread function is defined as

$$\text{LSF}(X) = \int_{-\infty}^{+\infty} h(x - \xi, y_0 - \eta) d\eta = \int_{-\infty}^{+\infty} \text{PSF}(x, y) dy, \quad (2.17)$$

where y_0 , is arbitrary. LSF and ERF are unique imaging characteristics of 1-D system corresponding to PSF for a 2-D system.

2.3.5 Transfer function

In imaging systems, imaging characteristics which are expressed in the spatial domain can also be expressed in the spatial frequency domain. The two dimensional Fourier transform $G(u,v)$ is defined by [40]

$$\begin{aligned} G(u, v) &= F[g(x, y)] \\ &= \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} g(x, y) \exp[-i2\pi(ux + vy)] dx dy, \end{aligned} \quad (2.18)$$

where F is the Fourier transform operator, and u and v are the spatial frequencies in the x and y dimensions. Each u, v point in the Fourier space corresponds to an elementary plane wave type of function in object space. The inverse Fourier transform is

$$\begin{aligned} g(x, y) &= F^{-1}[G(u, v)] \\ &= \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} G(u, v) \exp[i2\pi(ux + vy)] du dv. \end{aligned} \quad (2.19)$$

where F^{-1} is the inverse Fourier transform operator. Equation 2.19 shows that the 2-D spatial image $h(x,y)$ is described as a sum of 2-D spatial frequency components according to the Fourier theorem. In other words, the original function can be subject to spatial frequency filtering through various degradations in the system.

The convolution of two functions in space is represented by simply multiplying their frequency spectra expressed as

$$S(u, v)H(u, v) = F[s(x, y) * h(x, y)], \quad (2.20)$$

where $S(u, v)$ is the complex spatial frequency spectrum of the source distribution, and $H(u, v)$ is the spatial transfer function of the system in the frequency domain. In a similar manner, a Fourier transform function for 1-D system can be derived.

The absolute value of the transfer function is the modulation transfer function (MTF),

$$MTF = |H(u, v)|, \quad (2.21)$$

which relates the image and source distribution amplitude spectra by

$$|G(u, v)| = MTF(u, v)|S(u, v)|. \quad (2.22)$$

For an object which is described by a distribution of spatial frequency components, the image is the sum of the products of all the object frequency components times the corresponding MTF value.

In simple objects, $g(x, y)$ is usually separable in its coordinates as [40]

$$g(x, y) = g_x(x)g_y(y), \quad (2.23)$$

then

$$F\{g(x, y)\} = F\{g_x(x)\}F\{g_y(y)\}. \quad (2.24)$$

This is a convenient simplification in evaluating the transforms.

A variety of interesting relationships result when the function is separable in polar coordinates, as given by

$$g(r, \theta) = g_r(r)g_\theta(\theta). \quad (2.25)$$

The Fourier transform has the form of

$$\begin{aligned} F\{g(r, \theta)\} &= F\{g_r(r)\}F\{g_\theta(\theta)\} \\ &= \sum_{n=-\infty}^{+\infty} a_n (-i)^n e^{in\phi} H_n\{g_r(r)\}, \end{aligned} \quad (2.26)$$

where the θ has been decomposed into its angular harmonics as

$$F\{g_\theta(\theta)\} = \sum_{n=-\infty}^{+\infty} a_n (-i)^n e^{in\phi}. \quad (2.27)$$

$H_n\{.\}$ is the Hankel transform of order n , which is given by

$$H_n\{g_r(r)\} = 2\pi \int_0^\infty r g_r(r) J_n(2\pi r \rho) dr, \quad (2.28)$$

where, ρ and ϕ are the Fourier transform variables in polar coordinates.

Thus the complete transform is given by

$$\begin{aligned} G(\rho, \phi) &= F\{g_r(r)g_\theta(\theta)\} \\ &= 2\pi \sum_{n=-\infty}^{+\infty} a_n (-i)^n e^{in\phi} \int_0^\infty r g_r(r) J_n(2\pi r \rho) dr. \end{aligned} \quad (2.29)$$

In the important case of circular symmetry, such as in most of fiber optics, with no θ variations,

$$g(r, \theta) = g_r(r). \quad (2.30)$$

All the a_n values are zero except for a_0 which is unity. The resultant transfer function becomes

$$G(\rho, \phi) = G(\rho) = \int_0^\infty g_R(r) J_0(2\pi r \rho) 2\pi r dr, \quad (2.31)$$

where, $J_0(2\pi r \rho)$ is the Kernel of the Fourier-Bessel transform for the functions having circular symmetry and ρ is the radial spatial frequency variable. The Hankel transform of zero order is often referred to as the Fourier-Bessel transform.

2.3.6 Sampling

In digital imaging and image analysis, the signal is sampled by computer operations. The continuous function is presented by a finite array of samples. The conditions under which these samples completely determine the image function should be well defined.

The sample function is usually an array of delta functions. Consider a 2-D image $g(x,y)$ which is sampled by X in the x dimension and Y in the y dimension. The resultant sampled image $g_s(x,y)$ is [40]

$$g_s(x,y) = g(x,y) \text{comb}(x/X) \text{comb}(y/Y). \quad (2.32)$$

The comb functions are the array of delta functions separated by $1/X$ and $1/Y$ due to the finite sampling. The spectrum of the sampled image $g_s(x,y)$ is

$$\begin{aligned} G_s(u,v) &= F\{g_s(x,y)\} \\ &= G(u,v) ** XY \text{comb}(ux) \text{comb}(vy), \end{aligned} \quad (2.33)$$

where the 2-D convolution operation is abbreviated as

$$g ** h. \quad (2.34)$$

As shown in many textbooks [42] one can approach the band limitation as the image having no frequency components greater than half of the sampling frequencies, $1/(2X)$ and $1/(2Y)$. Any greater image

frequency components should be removed by filtering, otherwise the components will cause the overlap in the spectral islands and result in spatial aliasing.

2.4 ITS Monte Carlo Code

The integrated tiger series is a sophisticated Monte Carlo software package [28]. The code processes photon/electron transport, with or without the presence of macroscopic electric and magnetic fields. The ITS code was originally developed and is maintained by Sandia National Laboratories.

The ITS code is written in three dimensional geometry and treats continuous energy particle transport. Physical rigor is maximized by employing the best available cross sections and sampling distributions, and the complete physical model for describing the production and transport of the electron/photon cascades from 1.0 GeV down to 1.0 keV.

In the interactions of any particles, the simulation can simply select a distance to a next collision site, transport to the collision site, select a new energy and direction following the collision and then continue to a new collision cycle. In practice it is not true for electrons. Unlike the neutral particles that can be described as traveling in straight lines if there are no collision sites in the medium, the charged particles, such as electrons and positrons, continuously interact with the electric field. Therefore, it is difficult to define a collision although the concept of electron interaction is discrete collisions. Besides, unlike neutrons which

only require on the order of tens to hundreds of collisions to slow down from the keV to the eV range, electrons require hundreds of thousands collisions to slow down. Therefore, the direct simulation of collision by collision transport of electrons could be extremely computer time consuming.

The ITS code treats the coupled electron/photon radiation transport problems by the condensed history Monte Carlo method. The condensed history method is based on the concept that the electrons undergo many collisions and any one may be of little or no consequence to the energy and direction change; however, the cumulative effects of many collisions can be integrated and lead to an important change on both the direction and energy of the electron. The condensed history method condenses many collisions into a single, multi-collision computational event. This method relies on the theoretical multi-scatter distributions developed in the 1940s by Moliere [43] and Goudsmit [44], and is very efficient for electron transport calculations.

2.5 Photon Radiation Sources

Photon sources used in radiography are either isotope sources or x-ray sources generated by an x-ray tube or an accelerator. The sources from isotopes are monoenergetic while x-ray sources are spectrum sources.

The x-ray source model is a heavy metal bombarded by current electrons accelerated by a constant potential. As the electrons slow down

in the target both Bremsstrahlung and characteristic radiation are produced. Bremsstrahlung is emitted by the electrons as they decelerate in the nuclear field of the heavy metal atoms, the energy of the electron converted to photon energy up to its full kinetic energy. Because the energy given off with each photon-electron interaction varies, a spectrum of photons is generated with its maximum energy equal to the electron accelerating energy. Characteristic radiation arises from ionizing target atoms and the subsequent emission of radiation from electrons filling the empty positions. Since the energy levels of electrons are discrete according to their shell position, characteristic radiation contributed to discrete energy characteristics of the x-ray spectrum.

The x-ray energy spectrum is the ultimate measure of beam quality. It is not easy to measure directly and, therefore usually obtained analytically. D. Fry has modeled an x-ray generation from x-ray tube in his master thesis [45]. Also, a two parameter equivalent spectrum [EQSPEC] and later, a three parameter equivalent spectrum model has been developed.[46,47]

In our research, radiation sources from isotopes, x-ray tubes and linear accelerators are utilized to evaluate SFO image detectors. The ITS Monte Carte code has been used to simulate the generation of the x-ray spectrum produced by a linear accelerator. The conversion efficiency of the target and the spectral quality of the x-rays are also determined by the simulation.

2.6 Radiation Interaction

X-rays and γ -rays are electromagnetic radiations, whose wavelengths are many orders of magnitude shorter than those of visible photons. X-ray and γ -ray photons (photons in short afterwards) are absorbed or scattered in single events. A beam of these well collimated photons incident on matter undergoes a true exponential attenuation. The photons which penetrate the matter will have no interaction, while those which undergo single interactions are eliminated from the beam. In each case, the process is an all-or-nothing affair.

The primary interaction of photons with matter is the discrete collision with electrons. Some twenty five well-defined interaction processes contribute to the photon-electron cascade [48]. A photon can interact with matter by any one of these competing alternative mechanisms. The dominant photon interactions in the range of 10 keV-10 MeV are: photoelectric absorption, pair production, and Compton scattering [49]. Each of these interactions produce one or two (by pair production) energetic secondary electrons. These electrons then produce many thousands of ionization events as they slow down.

The fraction f of the incident quanta which are attenuated, i.e., undergo one of these interactions, in their passage through a matter of thickness t is

$$f = I(t) / I_0 = 1 - e^{-\int_0^{E_{\max}} \mu(E) t \cdot dE}, \quad (2.35)$$

where, I_0 is the intensity of incident photons, I is the intensity of I_0 after passing the thickness of t , and $\mu(E)$ is the photon linear attenuation coefficient at energy E .

$$\mu(E) = \tau(E) + \sigma(E) + \chi(E), \quad (2.36)$$

τ , σ and χ are the photoelectric, Compton scattering and pair production linear attenuation cross sections respectively. Each of these quantities depends on both the photon energy and the interaction matter. As shown in Figure 2.1, at low energies τ has the largest component, but it decreases rapidly with increasing the incident energy E , though with heavy elements it may still be important up to a few MeV. σ dominates at the intermediate energies. Pair production starts at energy of 1.02 MeV and increases steadily with increasing E .

It should be noted that within all but the very lightest nuclei the average binding energy per neutron or proton is between 7 and 9 MeV [50]. Thus a nucleus can absorb a high energy photon and emit a neutron or a proton as the photon energy exceeds the separation energy from the emitted nucleon. The maximum value of these (γ, n) or (γ, p) cross section is smaller than 5% of the total cross section of the same atom for Compton and Pair production interactions.

Figure 2.1 Relative Importance of the Three Major Types of γ -Ray Interaction

The lines show the values of Z and $h\nu$ for which the two neighboring effects are just equal. From Evans (1955) [8].

In the photoelectric effect, a photon interacts with an entire atom, resulting in the emission of a photoelectron, from the K shell or L shell of their atoms. Momentum is conserved by the recoil of the residual atom. Although the energy excess of the electron binding energy B_e is

distributed between the electron and the recoil atom, virtually all of that energy is carried as kinetic energy of the photoelectron, as

$$T_p = E - B_e. \quad (2.37)$$

In the process of Compton scattering, a photon of energy E interacts with an electron initially at rest. After the interaction, the photon has energy E' and moves at angle θ_s measured from its initial direction of the photon. The electron moves with kinetic energy T_{cs} at angle θ_e measured from the initial direction of the photon. From the principles of conservation of energy and linear momentum leads to the well known Compton formula,

$$T_{cs} = \frac{E}{1 + \left(E / m_0 c^2 \right) (1 - \cos \theta_s)}. \quad (2.38)$$

T_{cs} is from 0 to $\frac{E}{1 + 2E / m_0 c^2}$, depending the scattering angle θ_s .

When the incident photon energy E exceeds $2m_0c^2=1.02$ MeV, pair production becomes possible. In this process, the photon is completely absorbed and its energy is converted into the rest mass energy $2m_0c^2$ and the kinetic energy of an electron-positron pair

$$T_{pp} = T_{e-} + T_{e+} = E - 2m_0c^2. \quad (2.39)$$

At the end of its range the positron interacts with an electron, producing two annihilation radiation photons moving in opposite directions, each of energy 0.51 MeV.

Each of the three interaction processes yields secondary electrons whose kinetic energy T are related the initial photon E in different ways. These energetic secondary electrons in their passage traversing matter dissipate their energy in excitation and ionization, and scatter in collision with electrons and nuclei. Energy is also dissipated as Bremsstrahlung radiation. The excitation and ionization of the atoms of the scintillation material generate scintillating light and this is the basic working principle of scintillation detections.

Electrons may suffer appreciable deflection so that their path through the matter are tortuous. In practice, electrons transmitted against the thickness of the detector is known as range. The empirical equation to calculate the electron range is [51]

$$r(\text{mg} \cdot \text{cm}^{-2}) = 530E(\text{MeV}) - 106. \quad (2.40)$$

Equation 2.40 is applicable to any scintillation material.

The complexity arises, however, from the fact that not only the primary interaction yields secondary electrons, but the photon-electron cascades. The residual energy of the incident photons, which is not transferred to electrons or positrons at the primary interactions, is present in the matter as photons of reduced energy, x-rays of energy B_e ,

photons of energy $E - T_{cs}$, or annihilation quanta of energy m_0c^2 . These secondary photons will have a probability of interaction and convert their energy to electron energy. The electron energy will be added in producing a scintillation. By sequence of such multiple photon-electron transport, the incident photon may convert more energy to electron energy. However, these electrons will typically be of the primary beam and degrade the spatial resolution.

2.7 Luminescence

Luminescence is the phenomenon of emission of electromagnetic radiation in excess of thermal radiation [52]. The radiation is usually in the visible region of the spectrum. Excitation of the luminescent substance is prerequisite to luminescent emission, photoluminescence depends on excitation by electromagnetic radiation or photons; cathodoluminescence by energetic electrons or cathode rays. Luminescent emission can also be stimulated by chemical reaction, thermal excitation, and mechanical energy. In radiation detection, luminescence is generated by radiation particles and usually referred to scintillation. For gamma ray or x-ray detection, secondary electrons generated by photons will dissipate their energy in excitation and ionization scintillation atoms, which generate scintillating light.

The general phenomena of luminescence has been classified according to the duration of the emission following removal of excitation [53]. When excitation is removed, there is invariably an exponential

afterglow, whose rate is independent of excitation intensity and of temperature. This is often termed fluorescence. Frequently there is an additional component of the afterglow corresponding to the emission of longer wavelength light that decays more slowly. This component is phosphorescence. Delayed fluorescence results in the same emission spectrum as fluorescence but again is characterized by a much longer emission time following excitation.

Luminescence is observed with all phases of matter: gases, liquids and solids, both inorganic and organic. The luminescent inorganic solids are often called phosphors. The luminescence mechanism of most inorganic materials depends on the energy states determined by the crystal lattice. As shown in Figure 2.2, electrons have available only discrete bands of energy in materials classified as insulators or semiconductors. The lower band, called the valence band, represents these electrons that are essentially bound at lattice sites, whereas the conduction band represents those electrons have sufficient energy to be free to migrate throughout the crystal. There exists an intermediate band of energy called the forbidden band, in which electrons can never exists. In the pure crystal, the return of the electron to the valence band with the emission of a photon is an inefficient process. Furthermore, typical gap widths are such that the resulting photon would be of too high energy to lie in the visible range.

Figure 2.2 Energy Bands in Impurity Activated Crystal Phosphor

To enhance the probability of visible photon emission during the deexcitation process, small amounts of an impurity or structure defects are commonly added to inorganic materials. These imperfections are called activators, which create energy states within the forbidden gap through which the electron can deexcite back to the valence band. This transition serve as the basis of the luminescence process [54].

Activators produce local energy levels in the normally forbidden region between the conduction and valence bands. If these levels are

unoccupied, electrons moving in their vicinity may enter these centers.

The centers are of three main types [51]:

- (a) luminescence centers, in which the transition to the ground state is accompanied by photon emission;
- (b) quenching centers, in which radiationless thermal dissipation of excitation energy may occur;
- (c) traps, which have metastable levels from which the electrons may subsequently return to the conduction band by acquiring thermal energy from the lattice vibrations, or fall to the valence band by a radiationless transition. The energy level system for an impurity-activated crystal phosphor is shown schematically in Figure 2.3

The condition for luminescence emission and quenching of a center can be discussed in terms of a theoretical model proposed by Von Hippel (1936) and Seitz (1938,1939). The potential energies of the ground and excited electronic states of the luminescence center are plotted in Figure 2.3. The minima A and C correspond to the stable energy positions in the two states. At normal temperatures, the thermal vibrations cause displacement from the minimum potential energy positions corresponding to energies of the order of kT .

The direct absorption of a photon $h\nu$ by the center causes a transition from the ground to the excited state of the system as shown in the vertical line AB in the diagram. Immediately after this transition the system is not in a state of minimum potential energy, and so it moves from B to C and the excess vibrational energy is dissipated thermally to its neighbors. Then, the optical transition CD gives rise to the luminescence

emission $h\nu'$. After this transition occurs the center is in its ground state and returns from D to A by further thermal dissipation.

Figure 2.3 Potential Energy Diagram of Luminescence Center

The potential energy curves of the ground and excited states usually intersect or approach each other closely at some point F. A luminescence center in the point F can make a non-radiative transition to the point G of the ground state, and hence dissipate the excess energy thermally. This corresponds to the internal quenching process.

Figure 2.4 shows the origin of the absorption and emission spectra correspondent to the AB and CD transitions respectively. The emission spectrum is at lower energies (longer wavelengths) than the absorption spectrum. The overlap of the absorption and emission spectra, which occurs in some cases, depends on the relative positions of the potential energy curves of the ground and excited states. Usually the absorption and emission spectra of luminescence centers in inorganic phosphors are broad bands, characteristic of transitions disturbed by thermal motion and interaction with the crystal lattice.

Figure 2.4 Absorption and Emission Transitions

Fiber optics are made by SiO_2 glasses. Glass-forming materials in themselves do not luminesce, but the presence of suitable activators leads to numerous luminescent phenomena. Contrary to the case of crystalline solids, the lack of periodicity in the atomic arrangement leads to potential luminescence centers in a wide variety of energetically different environments within a given glass. In general, this leads to a broadening of the luminescence bands as compared to the bands of similar centers in crystals.

Glasses are excellent solvents. No matter what is put into a glass, at least in small amounts, it can be incorporated as part of the glass structure by rearrangement of the constituent atoms. Therefore, structural impurities do not exist in the same sense as in crystal.

The possibility exists of producing a wide variety of types of luminescence centers because the centers are in energetically different environments. Luminescence centers of glass consist of [55]:

- (a) "energetically isolated" activators, such as metal atoms, which are considered to exist as a metal vapor in the glass;
- (b) molecules or groups that possess such a highly covalent nature as to be considered molecular groups;
- (c) certain ions which participate as an integral part of the glass structure.

The luminescence centers of rare-earth glasses are ions. The ions in a specific quantum state will emit light when suitably activated. In the case of excitation of the outer electrons, such as that of the outer 3d electrons of the transition elements, the role of the host glass is important.

Figure 2.5 Energy Level Diagram of Terbium in Fused Silica
(data from Nelson et al., 1964)

2.8 Scintillating fiber optical image detectors

Scintillating fiber optics, as other detectors used in radiography, can be described by six factors [19].

- (1) Photon energy absorption efficiency, η_a , defined as,

However, in the rare-earth ions of glasses, the electronic transitions occur among the inner, shielded, 4f electrons. Both the absorption and emission bands are relatively sharp and are not greatly influenced by their surroundings. Therefore, changes in the composition and structure of the glass leads to no large changes in the luminescence.

Luminescence under ultraviolet excitation has been observed in glasses containing terbium, cerium, praseodymium, thulium, etc. Terbium glasses give a green luminescence [55]. With 0.5% Tb a large number of narrow bands occur, peaking at 401, 413, 417, 435, 457, 494, 520, 541, 546, 582, 590, 620, 654, and 679 nm.

The luminescence of Tb^{3+} in fused silica was studied by Nelson (1964). Under 253.7 nm radiation, Tb^{3+} was shown to emit the wavelengths from 377 nm to 653 nm. Two luminescent levels, $^5\text{D}_3$ and $^5\text{D}_4$ are found in the terbium ions which have lifetimes of 0.9 and 1.7 msec respectively. The energy level diagram obtained from their measurements is given in Figure 2.5. The bands observed for transitions from the $^5\text{D}_4$ level to the $^7\text{F}_6$ multiplet were called band system I; those from $^5\text{D}_3$ to $^7\text{F}_6$ were called band system II. Terbium by itself leads to an emission in band systems I and II when excited by 253.7 nm radiation. Although the total emitted intensity increases with terbium concentration, a larger proportion of the absorbed energy is transferred to band system II at the expense of band system I [55].

$$\eta_a = \text{absorbed photon energy} / \text{incident photon energy}. \quad (2.41)$$

A high η_a is preferred for two reasons. First the detector must stop most of the incident photon energy to realize high detector efficiency.

Secondly, a high η_a results in a higher interaction rate and, therefore, lower statistical noise.

(2) Light conversion efficiency, η_c , defined as,

$$\eta_c = \text{generated scintillation light energy} / \text{absorbed photon energy} \quad (2.42)$$

This is the parameter sometimes called quantum efficiency.

(3) Transmission efficiency, η_t , defined as,

$$\eta_t = \text{output scintillation light energy} / \text{generated scintillation light energy}. \quad (2.43)$$

The overall detector efficiency defined as the output energy divided by the incident energy, is the product of $\eta_a \cdot \eta_c \cdot \eta_t$.

A fiber has a core material surrounded by a cladding material having lower refractive index. Since the scintillation light is emitted isotropically, only the part of the scintillation light emitted in directions lying within the critical angle is trapped, transmitted, and detected (see Figure 2.6). The unguided light which propagates from fiber to fiber,

without carrying spatial information, is referred to as cross talk. The cross talk degrades the spatial resolution. Interstitial material called extra mural absorber (EMA) can be used to absorb some of the unguided light, and therefore reduce the cross talk [4]

Figure 2.6 Principle of Photon Detection of SFO

Attenuation of light guided by a fiber is composed of three factors: (a) the absorption loss due to self absorption and impurities, (b) the reflection loss because of imperfect reflective condition at the core/clad interface, and (c) Fresnel reflection loss at the fiber face.

(4) Emission spectrum defined as the wavelength spectrum of the generated light. The emission spectrum of the scintillation light peak should be a wavelength optimized for optical transmission and subsequent electronic detection. The emission and absorption spectrum of the

scintillation material should overlap as little as possible in order to minimize self absorption.

(5) Time constant defined as the decay time of the luminescence. In a real-time or scanning system where the images are acquired sequentially at high rates, the time constant, τ , for the scintillation transitions should be sufficiently short to avoid smearing the light generated from previous lines of the scan into the present one. The detector system can be characterized by a single time constant as

$$f = \frac{1}{2 \cdot \pi \cdot \tau}, \quad (2.44)$$

where f is the upper frequency in Hz at which the response is 70.7% of the middle band response.

(6) Spatial resolution defined as the spatial resolving capability of an electro-optical imaging system. The spatial resolution depends on the size of individual fibers, the cross talk between the fibers, incident x-ray energy and the geometry of the detector.

The photon energy absorption involves the conversion of the incident photon energy into the kinetic energy of the secondary electron energy. The detector absorption efficiency, η_a , can be, to a good approximation, expressed by

$$\eta_a = I(t) / I_0 = 1 - e^{-\sum_i \int_0^{E_{\max}} \mu_i(E) \cdot dE}. \quad (2.45)$$

This equation is the modification of Equation 2.35 in which $\mu_i(E)$ is the linear attenuation coefficient of the i th material of a SFO detector. The absorption effect is the sum of the absorption of each material.

Obviously, a thicker detector and/or a higher $\mu_i(E)$ will have a higher η_a . The interaction rate, N , is proportional to η_a . Since Poisson statistics governs the detection of radiation, the signal-to-noise ratio (SNR) is proportional to the square root of the interaction rate. That is,

$$\begin{aligned} \text{interaction rate } N &\propto \eta_a, \\ \text{signal / noise} &\propto \frac{N}{\sqrt{N}} = \sqrt{N}. \end{aligned} \tag{2.46}$$

Therefore, detectors with higher η_a results in a higher signal to noise ratio for a given exposure period.

During the slowing down of the secondary electrons in a scintillation material, atoms are ionized. Their electrons at the valence band are excited and produce luminescence, that is, the visible scintillation light of de-excitation. The exact nature of this process is quite complex [56], and therefore, the light is usually assumed to be emitted isotropically.

Light conversion efficiency has been defined to be the fraction of absorbed incident photon energy which is converted to scintillation light.

One would always prefer this efficiency to be as large as possible.

Unfortunately, there are other processes such as quenching which de-excite atoms without scintillation. When the scintillating light is generated in a fiber, only trapped light will be transmitted through the detector.

High output light intensity can be obtained only when fiber materials with high light transmission efficiency are used.

As a secondary electron absorbs energy from the incident photon, light is generated during the slowing down path of the secondary electron. The range of the secondary electron depends on the electron energy and the density of the material. The higher the energy and the lower the material density, the longer the range. For example, the range of a 4 Mev electron in Tb_2O_3 glass is approximately 0.2 mm, which in turn contributes to image degradation.

Secondary photon interaction occurring in different fibers can degrade the spatial resolution as shown in Figure 2.7. As the incident photon energy increases, the photon-electron interactions change from predominantly photoelectric to Compton scattering, and as a result the energy deposited due to secondary photon interaction increases which degrades the spatial resolution. However, the angular distribution of the secondary photons is not isotropic, and as the energy is increased further the distribution becomes more forward peaked as shown in Figure 2.8. For example, for 2 MeV photon interaction, the collision differential cross section in $10^{-27} \text{ cm}^2/\text{steradian}$ in different scattering angle θ are:

$\theta(^{\circ})$:	1	5	10	20	30	45	60	90	150
cross section:	79	77	70	50	33	18	11	6.8	4.7

Most of the secondary photons are scattered forward in an angle of $\theta < 45^{\circ}$. Assume a 2 cm thick detector and Compton interactions are happed at the detector entrance surface. If the secondary photons are scattered in a direction of $\theta = 45^{\circ}$, 90% photons will escape from the detector and only

10% photons will generate secondary interactions from the calculation by Equation 2.35. Therefore, for a high energy photon impulse input, the responding output is a peak arounded by a distribution with magnitude of 10% of the peak

Figure 2.7 Image Degradation due to Secondary Photon Scattering

Figure 2.8 The Number of Photons Scattered into a Unit Solid Angle, at a Mean Scattering Angle, θ

Overall, the spatial resolution depends on the size of the fiber, photon/electron transport and cross talk of stray light between the fibers. The size of the fiber determines the upper limit of the spatial resolution of the detector. The reduction of the spatial resolution by photon-electron cascades in the range of less than hundred micrometers will be discussed in Chapter V. However, if the extramural absorber (EMA) is insufficient to reduce the stray light, the cross talk will be significant. Spatial resolution limited by the cross talk is energy independent, but it is

thickness dependent. The scintillating light will spread out as occurs in phosphor screens. Light dispersed in SFO is less than seen in phosphor screens for two reasons. First, part of the light is guided by the fiber waveguide; and second, EMA will absorb some of the stray light. It is expected that selection of a fiber with a good EMA to reduce the stray light will be important.

The emission spectrum of the Tb_2O_3 -based SFO is concentrated primarily in a yellow-green line at 550 nm, which is an adequate wavelength for light transmission and detection. In addition, the absorption and emission spectrum are well separated. This means that in a Tb_2O_3 -based SFO, scintillating light will propagate through a long distance in a light waveguide or the material itself without strong attenuation [4].

In the process of energy absorption and scintillating light emission, an atom is excited to a higher energy level almost instantaneously, then decays to the ground state or other states of lower energy with emission of light called fluorescence. The process of fluorescence is the prompt emission of visible radiation from a substance following its excitation by some means. Some other processes that can also lead to emission of visible light, such as phosphorescence and delayed fluorescence will result in much slower time constants [57]. To be a good scintillator, a material should convert as large a fraction as possible of the incident radiation energy to prompt fluorescence. The light emitted can be in general modeled by,

$$I(t) = I_1 e^{-t/\tau_1} + I_2 e^{-t/\tau_2} + \dots = \sum_i I_i e^{-t/\tau_i} \quad (2.47)$$

where I_i are the intensities of the light and τ_i are their time constants. The decay time of the de-excitation determines the time constant of a scintillator. Most inorganic scintillators can be adequately characterized by a simple exponential, although more complex time behavior is sometimes observed [53].

2.7 Optical Model

Scintillation fibers are usually square fibers. The mode analysis for a step-index, square waveguide was developed by Marcatili [58]. Figure 2.9 shows a cross-sectional view of a single fiber. The cross-section is divided into 9 regions. Region 1 is the fiber core, and regions 2 through 9 are the cladding. Energy is transmitted in the core as well as in the cladding of the fiber. The combination of propagating and evanescent fields exist in the regions of 1 through 9. According to electromagnetic theory, there is the evanescent wave phenomena in the cladding, that is, an exponential loss of the energy in the fiber cladding as the distance from the fiber core is increased as shown in Figure 2.10. The attenuated waves in the direction of x and y with attenuation constants α_x and α_y determine the spatial rate of decay of the field in the cladding. Regions 6 through 9 are to be ignored because they have no common boundaries with the core

and the energy in these regions are assumed smaller than that of regions 2 through 5.

Figure 2.9 Cross Section of a Single Fiber

Figure 2.10 TE Modes in a Dielectric Slab Waveguide as a Function of X

The wave propagation direction can be expressed in terms of the various Kappas, κ , where

$$\bar{\kappa} = \kappa_x \hat{x} + \kappa_y \hat{y} + \kappa_z \hat{z}. \quad (2.48)$$

In the core, there are only propagation (sinusoidal) fields, while in the cladding there are both propagation and exponential (evanescent) fields. Therefore, the field in the cladding must be expressed by the decay constant $\bar{\alpha}$ besides $\bar{\kappa}$. The fields are mathematically expressed in the core as,

$$E \propto e^{\pm j\bar{\kappa} \cdot \bar{r}}, \quad (2.49)$$

and in the cladding,

$$E \propto e^{\pm j\bar{\kappa} \cdot \bar{r} + \bar{\alpha} \cdot \bar{r}}. \quad (2.50)$$

By solving the wave equation subject to the slab boundary conditions, the modes supported by the square waveguide can be divided into TE and TM modes [59]. The characteristic equations for the TE modes are

the even TE mode

$$\alpha_{x,y} = \kappa_{x,y} \cdot \tan(\kappa_{x,y} \cdot d), \quad (2.51)$$

and the odd TE mode

$$\alpha_{x,y} = -\kappa_{x,y} \cot(\kappa_{x,y} \cdot d), \quad (2.52)$$

where d is half of the fiber size. The relation between $\kappa_{x,y}$, $\alpha_{x,y}$ to the numerical aperture (NA) (see Equation 2.56) can be expressed as

$$\alpha_{x,y}^2 + \kappa_{x,y}^2 = k_0^2 \cdot (n_{\text{core}}^2 - n_{\text{clad}}^2) = k_0^2 \cdot \text{NA}^2, \quad (2.53)$$

where k_0 is the free space wave number, n_{core} and n_{clad} are the refractive indexes of the core and cladding respectively. The NA is used

to describe light gathering capability. Therefore if the difference between n_{core} and n_{clad} is large, more light can be transmitted through the fiber. This relation also defines a circle with radius R ,

$$R = k_0 \cdot d \cdot \sqrt{n_{\text{core}}^2 - n_{\text{clad}}^2}. \quad (2.54)$$

The characteristic equations for the even and odd TE modes can be written simply in graphical form. Figure 2.11 shows the characteristic equation diagram for κ_x . The intersections on the circle define the propagation conditions present for the TE modes. Similarly, a plot can be made for κ_y . It is easy to see that a large NA and d have a large circle, and the fiber is a high multimode waveguide.

Figure 2.11 Characteristic Equation Diagram of a Dielectric Slab Waveguide

After κ_x and κ_y have been determined, they can be displayed in a circle diagram as shown in Figure 2,12. A point which corresponds to both a κ_x and κ_y value represents a mode which will be supported by the fiber. Because the fiber is square, κ_x and κ_y have equal values, therefore, each dot represent both a TE and a TM mode. The number of modes, N_m is

$$N_m = N_x \cdot N_y$$

$$= \left(\frac{4d}{\lambda_0} \sqrt{n_{\text{core}}^2 - n_{\text{clad}}^2} \right)^2 = \frac{16d^2}{\lambda_0^2} (n_{\text{core}}^2 - n_{\text{clad}}^2). \quad (2.55)$$

Figure 2.12 Three Dimensional Mode Chart

where, sphere radius $\kappa = \sqrt{\kappa_x^2 + \kappa_y^2 + \kappa_z^2}$.

The scintillating fibers act as dielectric waveguides for the scintillating light. Not all of the scintillating light is coupled into the fiber optic modes, some power is always radiated at angles beyond the critical angle and into the leaky cladding modes. This situation is modeled by a vertical atomic electric dipole into a planar electric waveguide [60].

A three dimensional plot can also be made to show the propagation direction $\bar{\kappa}$ as in Figure 2.12. For any mode by a point in this plot, there is a definite κ_x and κ_y value, and thus a κ_z value. This κ_z can be found by locating a point P on the sphere directly above the mode point in the X-Y plane. The propagation angle can be found by connecting the origin to the P point, that is:

$$\theta_p = \sin^{-1} \left(\sqrt{\frac{\kappa_x^2 + \kappa_y^2}{\kappa}} \right). \quad (2.56)$$

The exit or the acceptance angle of the fiber is related by Snell's law in Equation 2.53,

$$\theta_e = \sin^{-1} \left(\sqrt{\frac{\kappa_x^2 + \kappa_y^2}{\kappa_0}} \right), \quad (2.57)$$

where, $\kappa_0 = \kappa / n_{\text{core}}$.

The power coupling into each mode is a function of incident angle. The higher order modes, which correspond to larger angles of incidence

are harder to couple into. The transmission efficiency of these modes is also lower than that of the lower order modes due to the larger propagation angles and distances. The energy coupling and distribution among these modes can be numerically calculated.

Ray analysis is based upon the property that light frequency fields in slowly varying media behave like plane waves. Its limitations are: a ray is not a field; it does not have an amplitude, a phase, or a spatial extent. However, for the application of SFO, the scintillating light is incoherent without phase information, the spatial pattern of the light output is not needed, and the calculation of energy distributed by modes is very difficult. Besides, for a highly multimode waveguide, as for case of a SFO (several hundred modes), most of the energy will be transmitted in the core and ray analysis based on Snell's law becomes reasonably accurate and simple. For these reasons ray methods have been used for propagation models [61].

As shown in Figure 2.13, the difference between the core and cladding indices determines the maximum angle that a ray can travel while being guided by total internal reflections. The maximum, or the critical angle ψ_c is simply,

$$\psi_c = \sin^{-1} \left(\frac{n_{\text{clad}}}{n_{\text{core}}} \right). \quad (2.58)$$

The ray remains at the same angle to the fiber axis as it travels in the fiber.

Figure 2.13 Light Transmission of Fiber Optics

At the input, the maximum angle of acceptance, θ_m is related to the numerical aperture by

$$NA = n_{\text{surr.}} \cdot \sin \theta_m = n_{\text{core}} \cdot \cos \psi_c = \sqrt{n_{\text{core}}^2 - n_{\text{clad}}^2}. \quad (2.59)$$

Because NA describes light gathering capability, the larger the NA, the larger the acceptance cone. The maximum output angle, by the principle of symmetry, can be calculated by the same equation as for the input angle. As the angle becomes larger, the reflection power becomes larger, which can be calculated using electromagnetic theory [45].

The general expression for the total light transmitted by a straight fiber waveguide can be developed by using an integral over the distribution of the light and the area of the fiber,

$$I_{\text{trans.}} = \iiint I(\theta, \varphi, z, \lambda) dA \cdot d\Omega \cdot dz \cdot d\lambda, \quad (2.60)$$

where $I(\theta, \phi, z, \lambda)$ is the distribution of the scintillating light in the medium of the fiber core, dA is the element of area in the cross section, $d\Omega$ is the element of solid angle, dz is the element of length along the fiber, and $d\lambda$ is the element of the wavelength.

A simplified expression for the transmission efficiency of SFO is developed as follows for the calculation of transmission efficiency in Chapter III. Further discussion on light transmission properties of SFO will be shown in Chapter V. Assume a scintillation fiber with length L is irradiated by a parallel x-ray beam as shown in Figure 2.6. The scintillation occurs at a point at distance z locally and the scintillation light is generated isotropically. Because the light is incoherent no interference and phase information is retained. As the scintillating light is emitted in a 2π solid angle, the scintillating rays whose angles of incidence on the fiber borders smaller than the critical angle Ψ_c will be reflected and transmitted. The amount of light which can be transmitted is approximately, $1 - n_{\text{core}} / n_{\text{clad}}$ [2]. Half of the scintillation light travels to the right end of the fiber to be detected. This end is coupled directly to a PMT with index matching grease. The other half of the light travels to the left end of the fiber, then is reflected back with Fresnel reflection loss occurring at the fiber face. The total distance this half of the light travels $z+L$. The attenuation of the light in the core is $e^{-1/\Lambda}$ where Λ is the total attenuation coefficient which accounts for reflection and absorption losses.

Using the above analyses, the transmission efficiency can be estimated as,

$$\begin{aligned}
 \eta_t &= \eta_{t1} + \eta_{t2} \\
 &= \frac{1}{2} \left(1 - \frac{n_{\text{core}}}{n_{\text{clad}}} \right) e^{-\frac{L}{2} \frac{1}{\Lambda}} \\
 &\quad + \frac{1}{2} \left(1 - \frac{n_{\text{core}}}{n_{\text{clad}}} \right) e^{-\frac{3L}{2} \frac{1}{\Lambda}} \left(1 - \frac{t_{\perp} + t_{\text{II}}}{2} \right),
 \end{aligned} \tag{2.61}$$

where η_{t1} is the transmission coefficient for light travelling to the right end of the fiber as shown in Figure 2.6, and η_{t2} is transmission coefficient for the light travelling to the left and reflected at the left face of the fiber, then back to the PMT at the right end. L is the thickness of the SFO. The mean distance for the light travelling to the right end is $1/2 * L$, and for the reflected light is $3/2 * L$, and [62],

$$\begin{aligned}
 t_{\perp} &= \frac{4n_{\text{core}} \cos \theta (n_0^2 - n_{\text{core}}^2 \sin^2 \theta)^{\frac{1}{2}}}{[n_{\text{core}} \cos \theta + (n_0^2 - n_{\text{core}}^2 \sin^2 \theta)^{\frac{1}{2}}]^2}, \\
 t_{\text{II}} &= \frac{4n_{\text{core}} n_0 \cos \theta (n_0^2 - n_{\text{core}}^2 \sin^2 \theta)^{\frac{1}{2}}}{[n_0^2 \cos \theta + n_{\text{core}} (n_0^2 - n_{\text{core}}^2 \sin^2 \theta)^{\frac{1}{2}}]^2}.
 \end{aligned} \tag{2.62}$$

Equation 2.59 is the Fresnel formula for transmissivity, where $t_{\perp}$ and $t_{\parallel}$ are the transmissivity of the light at the left face of the SFO, the paralleled component denoted by subscript $\parallel$ and the perpendicular component by $\perp$. n_0 is the refractive index of the medium outside the left face. θ is the average angle with respect to the fiber axis of light piped down the fiber. If there is no reflector at the entrance window of the detector, η_{t2} closes to zero. If the attenuation is neglected, the simplest estimation on transmission efficiency is obtained as

$$\eta_t = 1 / 2(1 - n_{\text{core}} / n_{\text{clad}}). \quad (2.63)$$

2.9 Summary

In this chapter, we have presented the general background theory which is related to the objective of this research. This background is used either in computer simulation or to understand the theoretical aspects of the research. Background presented are on fiber optics, image theory, radiation sources, detector physics and Monte Carlo technique.

CHAPTER III

SFO PROPERTY STUDIES

3.1 Introduction

This chapter discusses the general properties of Tb₂O₃ based scintillating fiber optics.

A comparison of different inorganic SFO is first presented for the preselection of the material. Based on this comparison Tb₂O₃ based scintillating fiber optics were chosen for the application of high energy photon imaging. The basic physical and optical properties of the Tb₂O₃ based scintillating fiber optics were analyzed. These properties are: detection efficiency, spatial resolution, time constant, spectral response, optical coupling, linearity, and radiation effect.

A brief theoretical description of corresponding properties of the SFO is followed by experimental results. The experimental results of the SFO offer design parameters for used as an image detector. The detector efficiency and spatial resolution data also provide information for the development of computer simulation and image analysis.

3.2 A comparison of inorganic scintillating fiber optics

There are several types of inorganic SFO that may find their application for photon image detection. They are made of either a fiber optic scintillator or coupled intimately to a fiber optic image plate. The objective of this section is to compare the characteristics of different inorganic scintillating fiber optics for application to high energy photon detection.

(1) Scintillating glass fiber optics

These SFO plates, using a terbium or cerium doped glass, are formed by fusion of 10-50 micron fibers with extra mural absorption (EMA) cladding. The plates are produced in different thicknesses, with or without a light reflector on the entrance surface. A detailed description of this type of SFO is described in the next section.

The manufacturers of scintillating glass fiber optics are: Synergistic Detector Designs [63], Schott Fiber Optics [64], Collimated Holes [90], and Levy Hill Laboratories Ltd [75].

(2) Intagliated plates

The intagliated or filled-fiber-hole plate is shown in Figure 3.1. A commercially available hollow glass fiber material with fully etched holes is deposited on a standard fiber optic plate. The holes are filled by implanting $\text{Gd}_2\text{O}_2\text{S}$ of about 2-3 micron mesh. Similar to scintillating glass fiber optics, the light is restricted to the hole because of the internal reflection of the walls and, in principle, the screen can be nearly as bright

as a settled screen while maintaining the resolution for thick detectors. However, when filling of the $\text{Gd}_2\text{O}_2\text{S}$ particles in smaller etched holes it is difficult to obtain an even deposition. Usually the holes have center-to-center spacing larger than 50-60 micros that in turn limits their spatial resolution.

Figure 3.1 Integrated Scintillating Fiber Optics

(3) Salt screens deposited on fiber plate

In this case, a $\text{Gd}_2\text{O}_2\text{S}$ salt screen, such as a piece of commercial Trimax 2, is coupled to a fiber plate by using silicon oil to reduce the loss between the two materials. This is a conventional phosphor screen plus a

fiber optic waveguide. The spatial resolution of this detector is equal to that of the phosphor screen.

(4) CsI extruded fiber plate

Extruded fiber of CsI(Tl) and CsI(pure) have been recently manufactured by Solon Technologies [65]. CsI(Tl) and CsI(pure) are both non-hygroscopic materials and do not require hermetic encapsulation. Both have densities of 4.51 gm/cm³ and are of high Z number; therefore they could be efficient for high energy photon detection.

Fibers of both round and square cross sections have been extruded in diameters ranging from 1-3 mm. There was no special effort to treat their surfaces and the fibers have no cladding. Some light is expected to be reflected at the surface of the fibers that may improve their spatial resolution. However, it is reported that when the fibers become thinner the efficiency becomes worse and the manufacture becomes difficult.

(5) CsI(Na) crystal plate

A 2-D CsI(Na) scintillating plate with high spatial resolution has been developed by growing CsI(Na) crystal on a special fiber plate with fiber size of 20 microns [14]. The spatial resolution obtained is as high as 45 microns. The quantum efficiency of the plate compared with the standard CsI(Na) crystal is estimated to be 0.4.

This technique has some exciting potential because the cladding material is air that has the lowest reflective index. The light collection zone will increase, and therefore, the corresponding transmission efficiency. However, the thickness of the detector is limited by 150

microns because of the difficulty in growing of the crystals on the 20 micron fibers. Therefore, the application is limited on low energy x-ray detection. No information on commercial product has been found.

For the preselection of our applications a simple comparison of the five types of detector materials is made as follows. Intagliated plates and CsI extruded fiber plate can be made thick to meet the requirement of high energy photon detection; however, the fibers are difficult to be made thin enough to obtain decent spatial resolution. The limiting spatial resolutions of these detectors are at the range of 1 mm. Salt screens deposited on fiber plates are basic conventional phosphor screens that cannot maintain both the efficiency and spatial resolution at high energy photon detection. CsI(Na) crystal plates can not be made thick so that their applications are limited for soft x-ray detections.

Inorganic scintillating glass fiber optics can be made thick as several millimeter and the fiber can be as thin as 10 microns [4]. They have an adequate efficiency and spatial resolution for higher energy photon detection. Cerium doped glass fiber plate has a better quantum (scintillating) efficiency than the terbium doped one for low photon energy detection, but decreases more rapidly with increasing energy because its low density and self absorption of cerium material. Terbium based scintillating fiber optics have a high Z number and high density, and therefore they are selected for our application as high photon energy, high spatial resolution image detectors.

3.3 Tb₂O₃-based SFO Array

Two types of Tb₂O₃ based SFO were experimentally evaluated, one was provided by Synergistic Detector Designs which we refer to as SDD, and second was provided by Schott Fiber Optics designated as LG9. The SDD plate is 25 mm*3 mm in cross section and 5 mm thick, while LG9 is 26 mm*3 mm and 6 mm thick. Selected mechanical and optical properties of these two SFO's are listed in the Table 1. The microscope images of fiber lattice are shown in Figure 3.2. From the magnified images, the LG9 shows a sharper core/cladding interface and larger core/clad area ratio.

Table 1 Some Properties of SFO

	SDD	LG9
index of refraction of the core	1.58	1.61
index of refraction of the clad	1.49	1.48
numerical aperture	0.52	0.63
primary wavelength of light emitted (nm)	550	550
the density of the core (g/cc)	3.0	2.6
reflection loss coefficient*	0.9993	0.9993
absorption coefficient (cm ⁻¹)**	0.025	0.025
fiber size (mm*mm)	12*12	15*15
SFO size (mm*mm*mm)	25*3*5	25*3*6

* data taken from [66], ** data taken from [67].

(a) SDD Scintillation Fiber Optics.

(b) LG9 Scintillation Fiber Optics.

Figure 3.2 The Microscope Image of Scintillating Fiber Optics

The chemical compositions of the Tb₂O₃ based SFO core are listed as follows,
 SDD: O, 25.78%; Si, 16.7%; Sr, 15.3%; Cs, 27.6%; Tb, 14.5% [68]

LG9: Tb_2O_3 , 6%; BaO , 23%; La_2O_3 , 20%; Li_2O , 2%; K_2O 7%; SiO_2 , 42% [69].

Both claddings are made by SiO_2 classes.

The measurements of efficiency, time constant, optical coupling, and spatial resolution, etc., are described in the following sections.

3.4 Efficiency Measurement

The determination of the efficiency of the SFO was made experimentally in combination with analysis using the ITS Monte Carlo code.

Experiments were performed with the two SFO plates, LG9 and SDD. Two 5 mm thick SDD plates were connected together to forming a SDD plate with 10 mm effective thickness, using index matching compound of refractive index 1.4648.

Monoenergetic gamma sources were used in the experiment. A monoenergetic isotopic source was selected rather than using an x-ray tube or an accelerator in order to eliminate the photon energy distribution. The isotope sources used are listed in Table 2.

Table 2 Isotope Source

No	Source	Energy	Activity
1	^{241}Am	60keV	500mCi
2	^{137}Cs	662keV	13.89mCi
3	^{60}Co	1.25MeV	2.31mCi.

* in each decay, Co60 emits two gamma rays of energy 1.17 and 1.33 Mev respectively.

The experimental apparatus for the efficiency measurement is shown in Figure 3.3. A collimated beam port was used to interface the gamma source to the SFO. Hence, buildup due to scattering is assumed to be negligible. The incident photon energy can be calculated by assuming that the gamma sources are isotropic sources. The total energy of gamma rays at distance r from the source in a small area dA is,

$$I = \frac{S}{4\pi r^2} dA. \quad (3.1)$$

The SFO was placed directly on the entrance of a calibrated photomultiplier assembly¹. The photomultiplier assembly includes a photomultiplier tube (PMT), a green filter and a shutter, and was enclosed in a black box to block stray light. The slide shutter is used to protect the

¹ The photomultiplier assembly was provided by professor S. A. Collins, Department of Electrical Engineering, OSU.

photomultiplier tube from exposure to room light when not in use. The green filter is a broadband Royln Additive Filter with a transmission band between approximately 500 and 800 nm. The transmission spectrum was chosen to correspond closely to the output spectrum of Tb₂O₃ based scintillating fiber optics. The photomultiplier assembly was calibrated by the department of electrical engineering.

Figure 3.3 Experimental Setup for Efficiency Measurements

The procedure for detector efficiency determination was as follows. First, the distance between the isotopic source and the SFO plate is carefully measured. The test room was then made as light tight as possible to reduce stray light. All the readings from the PMT were obtained from a digital voltmeter (DVM) and monitored by an oscilloscope. The gamma ray exposure measurements included four steps.

- (1) The gamma source and the PMT shutter were both closed, and the dark current of the PMT measured;
- (2) The gamma source was opened and the shutter was closed, and the response of gamma rays interaction with PMT measured;
- (3) The gamma source was closed, and the shutter was opened, and the response to the stray light measured;
- (4) The gamma source and the PMT shutter were both opened, and the response (including the dark current, the stray light, the gamma ray interaction and the scintillating light) measured

Therefore, the scintillating light output was obtained from the reading of step four subtracted from steps two and three. The output obtained from the digital voltage meter (DVM) have to convert to light power from the PMT calibration data [59].

The overall detector efficiency is,

$$\begin{aligned} \text{DE} &= \text{scintillation output energy/ photon input energy} \\ &= \eta_a \cdot \eta_c \cdot \eta_t \cdot \end{aligned} \quad (3.2)$$

The overall detector efficiency of a SFO depends on both the thicknesses of the detector and the incident photon energy. For the 60 KeV source, η_a is close to unity (see Table 3) which means most of the photon energy is absorbed and DE is close to the maximum. In this case, the DE for SDD is 0.23%, and for LG9 it is 0.30%. The DE of a 100 mg/cm² GOS screen is about 4% at 60 KeV. Therefore, the overall detector efficiency of the SFO is low compared with phosphor screens.

η_a was found by using the simulation ITS code and checked by using Equation 2.42. η_t is estimated by Equation 2.58. Then,

$$\eta_c = DE / \eta_a \cdot \eta_t. \quad (3.3)$$

Table 3 SFO Efficiency Parameters

LG9 scintillating fiber optics			
energy(MeV)	.06	0.66	1.25
η_a (%)	96	9.9	7.1
η_c (%)	8	9	12
η_t (%)	4	4	4
SDD scintillating fiber optics			
energy(MeV)	.06	0.66	1.25
η_a (%)	97	11	7.3
η_c (%)	6	7	10
η_t (%)	3	3	3

The results for η_a , η_c and η_t are summarized in Table 3. It is note that a careful model for the η_t calculation is developed in Chapter 5. The measurement using photons from a 1.25 Mev ^{60}Co source has a larger error because the available ^{60}Co source is weak and the signal/noise at the

PMT output is only about 0.5. The signal considered here is the scintillating light collected and amplified by the PMT. The noise includes the inherent PMT dark noise, the background noise and the PMT's response to the gamma-rays. The background noise is primarily due to the stray light which can be eliminated by good shielding. However, the gamma-rays can either pass through the SFO and enter the PMT window, or impinge upon the PMT directly. Although the wavelength of the gamma-rays is out of the sensitive wavelength of the PMT, the response of the PMT to gamma-rays is still significant. Therefore, the acceptable intensity of the gamma source should be large enough to obtain a good signal/noise ratio.

The major source of errors for the absolute η_c measurement are: (1). the spread in the emergent light from a fiber is not explicitly considered in the η_t calculation, but the experimental result for DE is based on the assumption that all of the light is collected; (2). the distance between the source and SFO is difficult to measure because the isotopes are located inside of containers; (3). the noise is large compared with the signal due to the poor efficiency of the SFO.

Using Equation 3.3, the η_c for SDD is shown to be 6%, and for LG9 it is 8%. η_a is energy and material dependent. To stop all the incident energy the SFO should be made thick enough to increase η_a . However, as the detector becomes thicker, more light will be absorbed and η_t will be reduced. There exists a tradeoff which should be optimized to obtain the best overall detector efficiency.

In radiography, two requirements must be satisfied. First, the number of interactions must be sufficient to obtain the specified contrast. Second, the scintillating light must be intense enough for detection. A SFO can always satisfy the first requirement because its high η_a results in high stopping power. The secondary requirement depends on the overall detector efficiency of the detector. As mentioned previously, the overall detector efficiency for the SFO is low; however, by adding an imaging intensifier in the SFO detector system, sufficient gain can be provided to attain the required contrast. It should be noted that signal-to-noise is a function of η_a . Therefore, the addition of an image intensifier has negligible effect on the signal-to-noise ratio.

3.4 Spatial Resolution

3.4.1 Methods

The Modulation Transfer Function (MTF) is a measure of the spatial frequency response characteristic of an image system, and is defined as the ratio of the modules of the Fourier transfer of the output to the modules of the Fourier transfer of the input. Having high spatial frequency represents its ability to resolve fine details.

There are four possible methods to determine the MTF of an x-ray system. As discussed below, each has advantages and disadvantages [3].

(1) Modulation of square wave method

In one dimensional system, MTF can be defined by

$$\text{MTF}(\nu) = |H(\nu)| = |O(\nu)| / |I(\nu)|, \quad (3.4)$$

in which $|O(\nu)|$ and $|I(\nu)|$ are the absolute values of the output and input in frequency domain, $|H(\nu)|$ is the modules of the transfer function. A physical interpretation of the MTF is shown in Figure 3.4. The input source distribution, $I(\nu)$ is a single sinusoidal spatial frequency. For example, $|I(\nu=1\text{mm}^{-1})|=1.0$. For a linear invariant system, the output image will also be a sinusoid function of the same frequency, but with a modulation of the system MTF as

$$|O(\nu=1)| = \text{MTF}(\nu=1) * |I(\nu=1)| = 0.7 * 1.0 = 0.7. \quad (3.5)$$

As shown in Figure 3.4, the modulation of the image is reduced to 70% of the input source modulation at $\nu=1\text{mm}^{-1}$. If a source has a distribution of spatial frequency components, the image is the sum of the products of all the source frequency components times the corresponding MTF.

Figure 3.4 Modulated Transfer Function

The modulation of square wave method is a direct application of the Equation 3.4. As shown in Figure 3.5.a, a photon impinges on a square wave type grating and creates a modulated pattern. The MTF can be obtained by measuring the contrast modulation in the pattern. It provides a simple determination of the MTF value at a certain frequency. The disadvantage is that the square wave may introduce irregularities [3]. Especially at higher frequencies, the square wave pattern is often combined with noise. Therefore, data analysis becomes difficult.

(2) Fourier transform of square wave pattern method

As shown in Figure 3.5.b. the experimental setup is the same as in the modulation of square wave method, but the data analysis is different. The image pattern of the grid is computed directly by a Fast Fourier Transform (FFT) and then the relative value of MTF is obtained from the calculation. The advantage is the irregularity is washed out due to integration over a large area of the grid. Also, the MTF is obtained by direct computer calculation from the measured image. The disadvantage is that the square wave pattern must be extended to a large number of cycles to obtain sufficient resolution in frequency space.

(3) Line spread function method

As shown in Figure 3.5.c, a beam of collimated photons passes a slit and is recorded by the system as a line spread function (LSF). The slit has to be assumed infinitely narrow to get an accurate result. The LSF is then calculated by Fourier transform to obtain the MTF as discussed in Chapter II.2. The advantage of this method is that the MTF can be provided in a single measurement. The disadvantage is the narrow slit experiment set up is difficult for high energy photon measurement.

(4) Edge response function method

To overcome the disadvantage of the LSF method, a sharp edge is used instead of the slit as shown in Figure 3.5.d. In principle, the edge can be made as thick as possible to stop photon particles.

The edge response function expresses spatial resolution performance of an image system in the spatial domain. The spatial resolution is presented by unsharpness. Unsharpness is defined as the width of blurring measured on the image of a sharp edge. The concept of

radiographic unsharpness is of great value in understanding radiographic detail sensitivity. The frequency resolution is the frequency-domain presentation of the unsharpness that is in the spatial domain. It is obvious that a smaller unsharpness value will have a better spatial resolution. The MTF is calculated from the responding edge response function. This method is good for high energy photon measurements. However, The edge response function (ERF) usually only contains a limited number of points that may results in a large error.

a. Modulation of square waves method

b. Fourier transform of square wave patterns method

Figure 3.5 MTF Determination Method

Figure 3.5 (continued)

c. Line spread function method

d. Edge response function method

The Edge response function method was chosen in our experiments to determine the spatial resolution of SFO detectors. Low energy photon measurements were carried out using industrial x-ray generators and the Scantech Diode Array Digital Radiography (DADR) system. The DADR system acquires the ERF data automatically and the spatial distance

between any two point is fixed at 50 microns. Therefore, the error due to the limited measurement points can be avoided. High energy photon measurements were carried out at Ohio State University James Authen Cancer Hospital using a Siemens linear accelerator and using Siemens linear accelerator at Ohio State University Hospital for high energy detection. The DADR system acquires the ERF data automatically and the spatial distance between any two point is fixed at 50 microns. Therefore, the error due to the limited measurement points can be avoided.

3.4.2 Experiment and Results

The objective of this experiment is to measure the inherent spatial resolution of the SFO detectors. The experimental setup is displayed in Figure 3.6. The SFO is the active element of an x-ray image camera. The camera consists of the SFO, a light intensifier, photodiode arrays, and an electronic amplifier. The optical image throughout the system is by contact -no lens is used. The accelerating voltage of the x-ray tube was 90 kV. A 90 kV acceleration voltage was selected so that a comparison could be made with the results measured with a scintillating screen. To obtain an accurate result, the experiment was set to minimize the stray light and their reflections.

Figure 3.6 Experimental Setup for Measuring for Low Energy Spatial Resolution

Figure 3.7 shows an edge response function produced experimentally by imaging the edge of a 0.020 inch thick tungsten plate. ERF is the spatial intensity distribution across the image of an infinitely sharp edge. An effective unsharpness is defined as the distance corresponding to 63% (1/e) change on signal contrast at the center of the ERF. However, the ERF showed in Figure 3.7 is the combination of SFO and the DADR system. In radiography, the system unsharpness is a combination effect of the individual unsharpness of channel components, usually expressed as the quadrature sum of the unsharpnesses of the individual components as

$$\text{system unsharpness} = \left[\sum (\text{unsharpness of each component})^2 \right]^{1/2} \quad (3.6)$$

The DADR system unsharpness for an optical input without the scintillating detector was determined experimentally to be 87 μm by ERF. Using this information the unsharpness of the SFO were calculated and listed in Table 4. In the calculation, we neglect the effect of the unsharpnesses of the light input and x-ray beam by assuming that both the light and x-ray are perfect parallel beam and the diffraction and refraction of the light are negligible.

Table 4 Unsharpness of the Detector at 90 keV (μm)

Tb ₂ O ₃ SFO	
LG9 (6mm thick)	12×10^1
SDD (5mm thick)	16×10^1
SDD (10mm thick)	17×10^1
Gd ₂ SO ₄ Phosphor Screen	
Trimax 4 (fine-30mg/cm ²)	20×10^1
Larex (regular-70mg/cm ²)	28×10^1
Kodak, and Rarex 120G (dense-120mg/cm ²)	80×10^1

Figure 3.7 Edge Spread Function

The modulation transfer function (MTF) was then calculated by using an ERF-to-MTF conversion routine [70]. It is common to specify the resolution capability of an electro-optical imaging system by its limiting resolution. Because the eye is normally considered to require an image contrast of about 3% (10% from some literature) in order to appreciate the light-to-dark transition in the bars, the limiting resolution in line-pairs/mm can be considered as the spatial frequency at which the MTF has fallen to three percent of the amplitude at zero frequency. It should be apparent that the limiting resolution does not fully describe the performance of a system. The shape of the MTF describes the spatial intensity distribution in the signal. Figure 3.8 shows the system MTF calculated digitally from the corresponding ERF as shown in Figure 3.7.

Figure 3.8 Modulated Transfer Function

Similar to the system ERF, the system MTF is composed of the MTF of a number of individual components. The components of DADR considered are: x-ray focal spot, SFO detector, image intensifier, photodiode array, and electronics. Because

$$MTF_{\text{system}} = \prod_i MTF_{\text{i component}} , \quad (3.7)$$

the detector MTF is determined by

$$\begin{aligned} MTF_{\text{detector}} &= MTF_{\text{system}} / \prod_i MTF_{\text{i component (except detector)}} \\ &= MTF_{\text{system}} / MTF_{\text{system without detector}} \end{aligned} \quad (3.8)$$

Figure 3.9 shows the detector MTFs calculated digitally by Equation 3.7. The corresponding limiting spatial resolutions of the SFO are determined to be about 8-9 lp/mm.

Figure 3.9 The MTF of SFO Detectors

As discussed in Section 3.2, the spatial resolution of the SFO is determined by: (1) the size of the fiber which determines the upper limits of the spatial resolution; (2) photon-electron cascades; (3) the cross talk between fibers which depends on how effective of the EMA is.

As expected, the spatial resolution of the SFO is better than those of phosphor screens. However, the unsharpness of the SFO determined by the experiment is less than $200\ \mu\text{m}$ while the diameter of the fiber is only $12\text{-}15\ \mu\text{m}$, and no significant scattering in this energy level. This means the dominant factor in determining the spatial resolution is the cross talk. LG9 has better resolution although its fiber size is larger than SDD because LG9 is a newer product and has a better EMA.

The spatial resolution as expected degrades as the thickness of SFO is increased from 5 mm to 10 mm because of the dispersion of unguided light, however, this degradation is negligible.

Measurements of spatial resolution for SFO for high energy photons was arranged in a relatively simple way. As shown in Figure 3.10, a tungsten edge, a 10 mm SDD SFO and a Kodak film were contacted tightly to avoid any disorder caused by a gap. The x-ray irradiated the edge and SFO that generated the image of edge response function, and then, was recorded by 100 TMAX Kodak film. The film has a spatial resolution above 50 lp/mm [91]. The object was irradiated at a distance about 1 meter to ensure a well-collimated x-ray beam.

Figure 3.10 Experimental Setup for High Energy Photon Detection

Two different x-ray sources were used. The first one is Siemens Mevatron KD-2 electron linear accelerator that is capable of producing x-ray energies of 6 and 18 MeV [92]. Another is Simens Stavilipan 1 that generates x-rays of energy below 250 keV.

Assuming the spatial resolution of the film is perfect compared with that of SFO, and the x-ray beam is a parallel beam., there are two main errors that degrade the spatial resolution measurement for high energy photons. One is caused by backscattering radiation and absorption unsharpness although the experiment configuration is arranged such that

the backscattering of x-rays is reduced to minimum [93]. The backscattering of x-rays are most important sources. They interact with the SFO and then generate the scintillating light with no spatial resolution information. Since these factors cannot be described quantitatively, in general, they modulate the intensity of the image forming radiation in an unknown manner, thus introducing errors into the measurement results. Second error is caused by the thicker edge. To stop 80% of high energy x-ray, a tungsten with a thickness of 28 mm was used. In that case, a very small obliquity will cause a sizeable error.

The Kodak film was exposed to a x-ray dose (tissue) of 1-2 rad, and then developed and printed as shown Figure 3.11. From the prints, the film is least sensitive to x-rays, therefore, the error due to the response of the film to the x-ray can be neglected. The edge response function on the film was read by the Scantech data acquisition system. In the film, the unsharpness of the SFO is about 300 μm for 200 keV x-rays and 420 μm for 6 MeV and 18 MeV x-rays. The inherent unsharpness of the 10 mm SDD SFO was experimentally determined to be 170 μm . The main error for the low energy experiment is due to the obliquity of the thick tungsten edge. Assuming the inherent unsharpness of the SFO is 170 μm , and the measurement system has degraded the unsharpness down to 300 μm in the low energy experiment. By Equation 3.6 the unsharpness of the measurement system is about 250 μm and assumes that this unsharpness is same for both low energy and high energy photons. After eliminating the system effect by using Equation 3.6, the unsharpness for high energy is about 340 μm that is about the unsharpness of a 400 mm regular phosphor

screen for low energy photon detection. Moreover, from the original prints, the image smears due to the backscattering can be clearly seen in the high energy image but not in the low energy image. This means that the unsharpness of 340 μm has included degradation resulting from backscattered radiation. The exact degradation of the backscattering is an unknown factor. However, even in this case the spatial resolution of a 10 mm SFO detector is equivalent to that of a 0.4 mm regular phosphor screen. This means the SFO will have a superior spatial resolution when a thick detector is necessary.

Two conclusions can be made for high energy photon detection. (1) The exact value of the spatial resolution is difficult to determine due to the experimental errors. The unsharpness of SFO is estimated to be 340 μm (about 4-5 lp/mm) including the backscattering. (2) In the high energy imaging, backscattering will reduce the spatial resolution. The design of shielding is very important, and can be completed by Monte Carlo method.

Figure 3.11 The ERF Developed from the Films

3.5 Time Constant

The time constant was measured under direct x-ray pulse excitation. The experiment was carried at Linac Laboratory, Department of Chemistry, The Ohio State University. The accelerator is a Varian linear accelerator, model V-7715-A [Manu of Varian Linear Accelerator, Model V-7715-A, Department of Chemistry, OSU]. The electron source is of pulsed radio frequency energy (3.0 GHz, S-band microwave magnetron rated for 2 megawatts peak pulse power). The acceleration process can be in single pulse mode in which the electron pulse width can be varied over the range from approximation 15 to 1400 nanoseconds. The electron to x-ray converter was either a lead plate which has higher Z number, or Tungsten plate which can withstand higher temperature. The thickness of the plate should thick enough to stop the majority of electrons but thin enough that most of the x-ray flux is able to penetrate the remainder of the plate. For high energy electron to x-ray conversion, the maximum x-ray flux is in the forward direction, therefore, the convertor is a plate. For low energy conversion, the converter is angled to increase the area for electron interaction and resulting in a smaller effective focal spot.

The experimental setup for time constant measurement is shown in Figure 3.12. A 4 Mev, 30 ns electron pulse was generated by a linear accelerator and converted to a x-ray pulse by a lead target. The selection of the target thickness was determined with the aid of the ITS code. The

generated x-ray intensity within $\theta=0-10^\circ$ is plotted in Figure 3.13. The optimized thickness of the Pb target is 0.8 mm for a 4 MeV electron input. The generated x-ray energy spectrum simulated by the ITS code is shown in Figure 3.14. The spectrum has a maximum energy equal to the electron energy, and an average energy of 0.664 Mev.

Figure 3.12 Experimental Setup for Time Constant Measurement

Figure 3.13 X-ray Density as A Function of Pb Target Thickness

Note: Normalized to One Incident Electron

Figure 3.14 X-ray Energy Spectrum

The experimental procedure has two parts. First, the output light from the SFO was blocked, and the signal recorded was the background noises, mainly due to the interaction between x-ray and the PMT. Then, the block was removed and the signal recorded is the sum of the light output and noise. The net light signal was obtained by subtracting the background noises. In all cases, the stray room light was eliminated.

The decay of the SFO fluorescence is displayed in Figure 3.15, and least squares fit to a two-exponential decays,

$$I(t) = I_0 (ae^{-t/\tau_1} + be^{-t/\tau_2}), \quad (3.9)$$

where τ_1 and τ_2 are the fluorescence decay time constants. Values obtained from curve fitting are given in Table 4. The characteristic time constant, τ is the time in which the light intensity is reduced to $1/e$.

Because $\tau_1 > \tau_2$, the first exponential decays slowly and the short time behavior is approximately,

$$I(t) = I_0 (a + be^{-t/\tau_2}), \quad t < \tau_1. \quad (3.10)$$

After a sufficient time, the second exponential decays to zero, and the long time behavior is determined by the first exponential,

$$I(t) = I_0 * a * e^{-t/\tau_1}, \quad t > \tau_2. \quad (3.11)$$

These results compare favorable with the fluorescence exponential decay time constant of Tb₂O₃-based SFO reported by Dr. Ruchti [4,5] where he measured a time constant of 3-10 ms when excited by 337 nm, 10 ns laser pulse.

Figure 3.15 Decay of Fluorescence of SFO Excited by X-ray Pulse

3.6 Optical Spectrum

The spectrum of Tb₂O₃ based SFO has been previously report by Richti [4], and later by Polichar [3]. There is a strong double peak near 550 nm with lesser peaks on either side. The broad distribution near 620 nm may be over emphasized due to the monochrometer rapidly changing response. This optical spectral matches most of the photodevices, such as fiber optical waveguides, image intensifiers, and photodetectors.

It should be noted that this scintillation spectral response, as well as the result from the time constant measurement match the report from Goldberg [52]. The wavelength emitted by 1000 ppm Tb^{+3} in fused silica is 541 and 548 nm, which correspond to the transitions from $^5\text{D}_4$ to $^7\text{F}_5$ level. The lifetime of the transition is 1.7 msec, which is very close to our measured response time of 1.9 msec.

3.7 Optical Coupling

The combination of a SFO and a fiber optic image waveguide comprises the photon image detector. The fiber optic waveguide is used so that the radiation sensitive electronic detector is located out of the direct path of the x-rays as displayed in Figure 3.16.

Figure 3.16 Fiber Optical Coupling Methods

The coupling efficiency between the square cross section of the SFO and cylindrical cross section fiber optic guide can be calculated by using either ray or modal analysis. Usually, the light attenuation inside the waveguide is small, and the major factor for calculation of the coupling efficiency is the overlap between the SFO output solid angle and the fiber waveguide input solid angle, which is shown in Figure 3.17. As the overlap decreases, that is, when using a more curved, or a more obliquely cut wave guide, the coupling efficiency will be reduced. A fiber optic image waveguide with a 30° obliquely cut angle with the fiber axis was tested. The wave guide is about 6 mm long and 4 mm*29 mm in cross section. The fibers of the waveguide are irregular in shape and approximately 6 to 7 mm by 10 mm. The coupling efficiency of the SDD scintillation fiber optics with the waveguide is 7% to 8% which is consistent with the measured results reported by Miller [59].

The combination of a Lanex phosphor screen with the same waveguide is also tested. The Lanex phosphor screen which is assumed to have a Lambertian angular distribution, couples approximately 10% of its output light into the waveguide.

Figure 3.17 Coupling Efficiency Visualization

3.8 Other Properties

Other properties which are required for an ideal scintillation material include:

- (1) the light yield should be proportional to deposited energy over as wide a range as possible;
- (2) the material should be of good optical quality and can be manufactured to a reasonable size;
- (3) radiation resistance

The linearity of Tb₂O₃-based SFO was reported by Polichar [3] in 1986. The Tb₂O₃-based SFO have good linearity. Dynamic range and saturation effect are not problems for these passive detectors.

The reflective index of SFO is near 1.5 which matches that of glass. The coupling of scintillation light to optical waveguides or image intensify devices should be efficient.

The SFO are widely available in various physical forms. The thickness of the plate can be made as thick as several centimeters, and the area of the plate can be large enough for most image application. However, the glass fibers are brittle and should avoid conditions of severe shock and vibration.

The current understanding of the radiation damage process in fiber optics is determined by performing a literature review. Pure or Doped silica comprises the core and cladding of all inorganic fiber optics. Undoped fibers have been presented of superior performance in radiation fields for high doses because of the radiation hardness of pure SiO₂ [94].

The concentration and nature of the dopants used in doped fibers determines the fiber's response to radiation [76]. Scintillating fiber optics are doped fibers. The dopants generate the scintillating light and raise silica's index of refraction. General speaking, inorganic scintillating crystals and glass have a dose limit around 10⁵ R [77]. Both thermal annealing and photobleaching are absorbed in all kinds of fiber optics which may significantly improve the life time of scintillating fiber optics detectors.

3.9 Conclusion

In this chapter, experimental methods and results were presented. The methodology was based on the analysis presented in Chapter II

Two types of Tb₂O₃ based SFO were experimentally evaluated using monoenergetic gamma sources, commercial x-ray generators and high energy electron accelerators used to generate high energy x-rays. The efficiency, spatial resolution, time constant, coupling efficiency and other properties were analyzed, measured, and discussed. The results from this chapter can be used to assess the practical feasibility of SFO as imaging detectors.

In the present form, the overall detector efficiency of a Tb₂O₃-based SFO is low due to the poor light conversion efficiency (η_c) and low light transmission efficiency (η_t). Both, however, can be improved by trying different scintillating materials. A fiber with a high core refractive index and low cladding refractive index is preferred to obtain good η_t and high stopping power. The transmission efficiency can be nearly doubled if a thin mirror transparent to x-rays is placed in front of the SFO to reflect the fraction of scintillating light emitted backward. However, η_t can never match with that observed in phosphor screens because only guided light can be transmitted. The absorption efficiency (η_a) of SFO can be maximized by adjusting the thickness of the detector, therefore, the interaction rate required to achieve design contrasts can always be satisfied. Because light generated by these

interactions is not sufficient for detection, an imaging intensifier is used to amplify the light.

Image blurring occurs in a phosphor screen because the scintillating light generated by x or gamma ray interactions with a scintillator is emitted isotropically. The SFO, on the other hand, is expected to exhibit high spatial resolution because of its special structure. The spatial resolution of the SFO was tested with low and high energy photons, and studied by both the ERF and the MTF. The experiment in low energy shows that the SFO has a fine spatial resolution of 8-9 lp/mm which is better than traditional phosphor screens and the spatial resolution is almost independent of its thickness. Although the exact value of high energy spatial resolution of the SFO cannot be determined due to experimental errors, the unsharpness of a 10 mm SDD SFO is estimated to be 340 μm (about 4-5 lp/mm) including the strong effect of backscattering which can be eliminated by shielding. It is expected that SFO detectors can be used at high energy radiography while maintaining good spatial resolution.

The time constant of the SFO is in the order of milliseconds which is not as fast as some other scintillating materials. In a real time system, the image detector should have as little memory as possible so that the light image at any instant reflects the incident x-ray image. The time constant of the SFO must be less than the time to acquire one pixel signal.

The Tb_2O_3 scintillator has well separated excited and emission spectra so that the self-absorption is small. The emission spectrum of the

light peak is at about 550 nm, hence the coupling of SFO with fiber optic waveguide or photodiode array should not be a problem.

The major defect of SFO at the moment is the poor quantum and transmission efficiency, but the study in Chapter 4 will show that the SFO combined with a light amplifier is feasible for use in radiographic systems. The images from the SFO-light amplifier based DADR system indicate that the SFO is superior to phosphor screens. SFO should be of value for nondestructive testing, medical imaging and high energy physics applications.

CHAPTER IV

The APPLICATION OF SCINTILLATING FIBER OPTICS IN DADR SYSTEM

4.1 Introduction

The objective of this research was to evaluate the use of scintillating fiber optics (SFO) to replace traditional phosphor screens as the imaging detector in a digital radiographic system. X-ray image quality is a function of two basic factors: number of photons detected per unit area and the spreading of the x-ray image inherent in the detection process. There exists a trade-off in the phosphor screen thickness for a given scintillator. An increase in the thickness (or screen weight) increases the signal-to-noise ratio (SNR), and also the detection efficiency. However, spatial resolution is degraded due to the increase of lateral spreading of the light with increasing the detector thickness. The result, however, is reduced spatial resolution. The advantage of SFO compared to phosphor screens is that the SFO can be made thicker. SFO exhibits superior SNR and x-ray detection efficiency, with minimal loss of spatial resolution through lateral light dispersion.

A SFO-based detector was incorporated into a digital radiographic system. The system is based on the patented Diode Array Digital Radiography (DADR) technology. For the purpose of evaluating the quality of SFO as an image detector in a DADR system, two types of Tb₂O₃-based SFO, LG9 and SDD, were experimentally evaluated. Two pieces of the SDD plate were available and were tested individually and also stacked (and optically coupled) into a detector 10 mm in thickness.

System quality was evaluated based on five criteria: spatial resolution, signal-to-noise ratio (SNR), large-area contrast detectability, responsivity, and inspection task performance. X-ray energies used to test SFO performance ranged from 90 to 300 KV_p. The experimental results were compared with those obtained with three commercial gadolinium oxysulfide (Gd₂O₂SO₄:Tb) phosphor screens. The phosphor screens used in these experiments were Trimax 4 (fine — 30 mg/cm²) by 3M, Lanex Regular (regular — 70 mg/cm²) manufactured by Kodak, and Rarex 120G (dense — 120 mg/cm²) obtained from Optonix. The experimental results indicate that SFO detectors are comparable to phosphor screens. Some of the properties are superior to those observed with phosphor screens. For high-energy detection, SFO is expected to be more efficient and exhibit superior performance.

4.2 Digital Radiography and DADR system

There are three types of digital radiographic systems based on different scanning methods [45]. In area digital radiographic systems, the

entire image area is exposed to an area x-ray beam at the same time with the detector covering the entire area. This is the only true real time digital x-ray imaging method. In point detection systems, a single detector and pencil beam of x-rays scans the entire imaging point by point to build up the total image. In line scanning systems, the image is built up one line at a time by sweeping a fan beam of x-rays across one dimension of the imaging area with a narrow detector covering the length of the other dimension.

Line scan digital radiographic systems take advantage of the scatter rejection of the detector geometry and electronic detectors to greatly improve the dynamic range and low contrast detectability of their images. Radiographic images are shadows of the variations in the specimen density and atomic number. X-ray images are not only produced by x-ray absorption, but also affected by Compton scattering and pair production. The broadly spread scattering will reduce the effective contrast of the incident x-ray distribution. Line scan technique permit the use of a collimated radial fan beam. The detector area is equal to the fan beam area cutting through the specimen. The width of the field is kept directly below the beam as the specimen through the beam and only a single narrow line of the part is illuminated with the radiation. The radiation scattered from the specimen under study is not detected. Therefore the usual background scatterer noise which degrades contrast sensitivity and spatial resolution is eliminated and a wider dynamic range of x-ray detectors is available at the detector. The line scan technique, however, has some disadvantages. Line images are integrated as the object is

scanned making the process necessarily slower than an area system. Also, the use of x-rays is inefficient due to the collimation of an area x-ray beam into a fan beam.

Diode Array Digital Radiography (DADR) is a method of radiographic imaging employing the line scan technique. DADR was invented at the University of Pittsburgh by a team lead by Sashin [71]. The intent was to develop a radiographic system which provides an image quality equal to or exceeding that of conventional film-screen radiography at a lower dose to the patient. Using the DADR technique, Scantech Corporation has developed systems for industrial Non-Destructive Testing (NDT) [72].

Figure 4.1 is a diagram of the DADR system. A linear x-ray camera is mounted beneath a shuttle table such that the x-ray beam is aligned with the image detector. The x-ray beam is slightly wider than the detector aperture so that the beam is collimated to its final detected width. The aperture is made from a dense material such as tungsten to shield the detector from scatter. The fan beam slice through the specimen and the moderate x-ray beam interacts with a narrow strip of scintillation detector to produce a visible light line image. The line image is optically amplified by an image intensifier and, then optically coupled to linear self-scanning photodiode arrays. The photodiodes convert the continuous, spatially distributed light signal into spatially discrete electrical charges by integrating their photocurrents over a specified period. Following the line integration period, the photodiodes in an array are read out sequentially to low noise preamplifiers to form a row of pixel

signals making up on line of the image. The image is then processed, amplified and stored in computer memory. In DADR system, the specimen is linearly translated through the fan beam and an image is acquired on a line-by-line basis.

Figure 4.1 DADR System

4.3 System Characterization

System image quality was evaluated with five characterization criteria.

(1) Spatial resolution is the spatial response characteristic of the system.

The system spatial resolution can be evaluated by spatial response or spatial frequency response as discussed in Chapter III.

(2) SNR is defined as signal level divided by one standard deviation of total system noise, typically analyzed as a function of another variable, such as object thickness.

(3) Large area contrast detectability is the minimum contrast which may be viewed on the system.

(4) Responsivity is a measure of the system sensitivity. The relative detector efficiency can be evaluated from the system response.

(5). Task of Inspection is the ability of the system to image actual defects in NDT.

The first four characteristics provide data for the design of the system and improvement of image quality. The last one is important in establishing functional equivalence of the image system. The functionally equivalent image means the desired level of inspection stringency is achieved. For example, the spatial resolution of the DADR system is not as good as that of fine grain film (>10 lp/mm). But in many applications, an image functionally equivalent to a film image may be produced with a DADR system [72]. For those cases, when the high spatial resolution is

not necessary, DADR may produce an image as good as or better than that recorded on films.

4.4 Experimental Results

The experiments involved acquiring images using the Scantech DADR prototype system. Measurements of performance for both scintillating phosphor screens and SFO plates were obtained. The radiographic technique and system gains were separately optimized for the screen experimental set and for the SFO experimental set. Pixel integration time and aft mask (scan-direction) aperture were set as part of the radiographic technique. Image intensifier gain and electronic amplifier gain were adjusted to provide adequate signal level input to the A/D converter for each test. The optimization of the x-ray technique was conducted to provide the best possible images of a commercial electronic sample/hold module and a radiographically-equivalent ASTM standard penetrometer. The technique selected for the SFO imaging work was 90 kVp and 12 mA, with an aft mask aperture setting of 0.46 mm and a pixel integration time of 200 mS. The technique of 45 kVp and 22 mA, with aft mask aperture equal to 0.25 mm and pixel integration time of 67 mS was used for the phosphor screen tests.

Lower kVp and higher current setting was necessary for tests using the phosphor screen in order to meet the ASTM standard. The settings were constrained by this goal and limited by the x-ray generator. At slightly higher kVp techniques, the contrast sensitivity of the phosphor

screens was noticeably reduced. Further increase in kVp for testing of either phosphor screens or SFO requires a corresponding reduction in tube current due to generator power limitations.

One exception to this is the system responsivity tests. These tests varied the technique as needed to establish the desired data points. The system gains, integration time, and aft mask aperture were held constant for all these tests, in order that the data might be directly compared.

The spatial resolution tests were performed with the optimized kilovoltage for the scintillator type, but lower tube current was used. This change was adjusted as needed to limit the input signal level to reasonable level at the A/D input. The lower tube current is required to avoid the effects of large detector response in the region where no object specimen is present.

The following subsections describe the experimental results obtained for spatial resolution, SNR, large-area contrast detectability, responsivity, and inspection task performance for the SFO detectors. In each case, the results are compared with those obtained for phosphor screens. The results from tests using different techniques and gain settings may be directly compared for spatial resolution and inspection task performance. In the cases of SNR and large-area contrast detectability, the raw data presented reflect differences in the detector performance and in system settings. For this reason, a second set of graphs is presented in which these effects have been normalized, assuming Poisson statistical effects on the measured signal and noise levels.

4.4.1 Spatial Resolution

In the evaluation of the spatial resolution of the SFO-based system, an edge response function (ERF) was first measured, and the modulation transfer function (MTF) was then calculated by using an ERF-to-MTF conversion routine [70]. The system ability to resolve fine details depends on both the shape and the limiting resolution of the MTF.

Due to the slit-scan nature of the line scan system, the MTF in the scan direction and in the detector direction are different. In the scan direction, aperture size defines the frequency response. However, in the detector direction, frequency is primary limited by the scintillating detectors. For the purpose of detector evaluation, only the MTF in the detector direction is examined.

The system spatial resolution based on different scintillating detectors are shown in Figures 4.2 and 4.3. The MTFs are also checked by a standard lead foil resolution pattern (Nuclear Associate Model 0.7-553, 0.25 to 10.0 lp/mm resolution).

Figure 4.2 System ERF

Figure 4.3 System MTF at 90 keV

The MTF of three different weights of gadolinium oxysulfide screens were used for comparison. Higher screen weights yield poorer spatial frequency response but are more efficient detectors. The experimental results show that the limiting spatial resolution of the SFO systems is 8-9 line-pairs/mm (lp/mm), which is superior to the 2-6 lp/mm resolution of the phosphor screen systems. (see Figure 4.3) The experiments also show that MTF of SFO has no significant degradation as the x-ray energy was changed from 90 to 300 KV_p and the thickness of SFO was doubled, as shown in Figures 4.3 and 4.4.

As mentioned in Chapter III, MTF of the system is composed of the MTF of a number of individual components. The relevant components of the DADR system are: x-ray tube focal spot, scintillator detector, image intensifier, photodiode array, and electronics. The system limiting resolution without a scintillating detector or image intensifier was experimentally found to be approximately 10 lp/mm. This was measured by using the ERF technique with visible light and confirmed by measuring the line-spread function (LSF) as well. The Nyquist cutoff frequency, determined by the diode spacing in the photodiode array, is also 10 lp/mm (as shown in Figure 4.3, the curve of perfect detector). From manufacturer's data, the image intensifier is known to have negligible effect on system MTF. Because the change among scintillator detectors produced a wide variation in MTF, it is concluded that the detector, rather than the focal spot or other factors, dominates the system resolution performance in the range of interest. Therefore, the improvement in MTF by use of SFO is an important design consideration.

Figure 4.4 System MTF at Different Energies

4.4.2 Signal-to-Noise Ratio

SNR is a significant factor in determining the usable information content of the image, primarily the ability of the system of discriminate contrast differences in the object. Quantum noise, which is the noise due to statistical variation in x-ray detection, is the primary noise source in x-ray radiographic systems. Therefore, the scintillating detector is the primary factor in determining SNR.

SNR has been measured by computing the mean and standard deviation of pixel values of image regions corresponding to the scan of individual steps of an aluminum step wedge. Measured data are plotted in

Figure 4.5. The results indicate that the SNR of SFO is better than that exhibited by phosphor screens because of the high stopping power, or absorption efficiency, of the SFO. The dense phosphor screen has a greater SNR but exhibits reduced spatial resolution as was shown in Figure 4.3. The SFO image detector, on the other hand, largely eliminates this trade-off. The x-ray imaging detector is the primary factor limiting the spatial resolution and SNR of an x-ray radiographic system; both these quantities are improved by use of SFO.

Figure 4.5 SNR vs Aluminum Thickness

As there are differences in the aft mask aperture, pixel integration time, and x-ray generator/tube current between the SFO and phosphor screen tests, Figure 4.6 shows the SNR data with the SFO data scaled to account for these differences. The assumption is made that the effect on SNR is approximately that of Poisson statistics, and that these three factors linearly affect N , the number of detected photons. The aft mask aperture linearly increases the sensing area, and the pixel integration time likewise increases the total number of photons measured. Tube current is also assumed to be linearly related to the number of photons produced by the x-ray source. The three factors were treated as ratios which could be multiplied together to determine a normalized N value. The effect on SNR is then $\sqrt{N}$. Following this reasoning, the SNR data for the SFO were scaled by the factor 0.62 for comparison with the phosphor screen SNR data on the normalized graph of Figure 4.6. As show, SFO performance still yields a greater SNR than for phosphor screens.

Figure 4.6 Normalized SNR vs Aluminum Thickness

The measured SNR data are approximately 1.0 to 1.5 times the amount predicted from Poisson statistics due to quantum noise. This indicates that there is some effect present other than x-ray quantum noise, or may be some due to some systematic experimental error. It has been hypothesized that this may be due to one or both of two effects. The first is that the data include corrections for offset due to detector diode dark current and corrections for gain variation from channel to channel in the detector array. These corrections are essential to the imaging system to produce high-quality flat-field images. The second is that summation of x-ray photons over an entire spectrum (including spectral effects of x-ray

generation, material interaction in the object specimen, and interaction in the scintillation detector) may not yield a strict Poisson statistical distribution in the final data.

The general expression of SNR for a system with q successive stages is [40]

$$\text{SNR} \propto \frac{\sqrt{\eta_a N}}{\sqrt{1 + \sum_{i=1}^q \left(1 / \prod_{j=1}^i g_j \right)}}, \quad (4.1)$$

where η_a is the fraction of incident photon energy which is absorbed by the detector, N is the number of photons per pixel element impinging on the detector, and g_j is the gain of j th stage of the system. If every product term, $g_1 g_2 g_3, \dots$, is appreciably greater than unity, the SNR will be dominated by the captured number of x-ray photons per pixel. In general, this is a desirable since it ensures that SNR has been maximized.

In a SFO-based image system, g_1 is the gain of the scintillating detector, which is the average number of light photons that can be transmitted through the SFO per x-ray photon. If a 100 keV x-ray photon produces electrons by the photoelectric effect, 6 keV of light photon energy is generated because the light conversion efficiency of Te_2O_3 -based SFO is about 6%. The SFO detector has a transmission efficiency of 3%; thus, the output light is about 180 eV. This corresponds to approximately 75 green photons (Tb_2O_3 emits mainly at 550 nm) and

results in a g_1 of 75. Therefore, this stage does not significantly change the resultant SNR. The SNR degradation contributed by the next stage is the image intensifier. In this device, light from the scintillating detector is amplified before utilization. The active conversion material of the image intensifier has a quantum efficiency of about 10%, so that one electron is emitted for about every 10 light photons. Thus, $g_2=0.1$. This provides a g_1g_2 product of $75 \times 0.1 = 7.5$, which results in about 7% degradation of the performance since the denominator of the expression is equal to 1.07.

4.4.3 Large-Area Contrast Detectability

Large-area contrast detectability corresponds to the minimum contrast which may be detected with the system. The minimum contrast is the smallest $\Delta T / T$ which may be observed. In this case, ΔT is the base thickness of the object and ΔT is some small change from that base value.

The large-area minimum resolvable thickness was calculated by a combination of the signal-vs-thickness and SNR-vs-thickness curves. It is assumed that one standard deviation (1σ) of the noise is the minimum discernable signal change. Dividing σ by the slope of the signal vs thickness curve (dS/dT) gives the minimum discernable thickness change. Doing this over the system latitude (measured in base material thickness and using fixed radiographic parameters, gain settings, integration time, and aft mask aperture) generates a minimum detectable thickness curve. The results are shown in Figure 4.7.

As in the case of SNR, another graph was generated in which the relative effects of scaling the aft mask aperture, pixel integration time,

and x-ray generator/tube current. The ratios of these parameters between the two test cases were presumed to linearly scale the number of photons detected, N . Hence, the large-area contrast detectability would be affected as $\sqrt{N}$, assuming Poisson statistics. Following this line of reasoning, the SFO data were scaled by the factor 1.61 for the normalized graph of Figure 4.8. From these analyses, it is concluded that large-area contrast detectability, or minimum detectable thickness, is comparable for SFO and phosphor screens.

Figure 4.7 Minimum Detectable Thickness

Figure 4.8 Normalized Minimum Detectable Thickness

4.4.4 Responsivity

Responsivity is defined as volts of amplifier output per mA of x-ray generator/tube current at a given material thickness, as a function of another variable, such as x-ray generator/tube kilovoltage. This is a measure of the system sensitivity. The goal is to detect sufficiently many x-ray photons to statistically resolve the desired features and to have a detectable signal level.

The measured data points are shown in Figure 4.9, along with the results from phosphor screens. The same electronic and image intensifier

gain settings, aft mask aperture, and pixel integration time were used for all of these measurements. The system sensitivity of the SFO is lower than that exhibited by phosphor screens at x-ray energy in the range of approximately 100-200 KV_p. This is the consequence of the low overall detector efficiency of SFO relative to that of phosphor screens in this energy range. In the measured digital system, the coupling fiber optics has been removed and a light intensification stage has been added to obtain improved system sensitivity. By adjusting the gain of the light amplifier and also the electronic amplifier, an adequately large signal output for a feasible x-ray dose input was obtained. The drawback of inclusion of an image intensifier is the added complexity, cost, an additional negligible MTF component and a modest reduction in SNR as was shown previously.

Figure 4.9 System Responsivity

4.4.5 Inspection Task Performance

Obviously, the final criterion for the success or failure of the SFO imaging detector is its ability to perform in some specific application. In the present case, this application is the inspection of two types of electronic assemblies, hybrid modules and standard circuit boards. One of the assemblies is a sample/hold module which is approximately radiographically equivalent to the ASTM 801E #4 penetrameter [73]. The image obtained met this standard, detecting all 6 lead particles and all 6 tungsten wires after the system radiographic technique, gain settings, integration time, and aft mask aperture were optimized. The finest wire detected was 12.7 μm diameter, and the smallest particle was nominally 50.8 μm in diameter. In addition, some fine details were identified in a standard circuit board which were not observed by using phosphor screens, and 6 μm tungsten wires were routinely visible [74].

4.5 Conclusion

A comparative test using SFO as an imaging detector in a DADR system has been completed. System quality was evaluated with five characterization criteria: spatial resolution, signal-to-noise ratio (SNR), large-area contrast detectability, responsivity, and inspection task performance. It is concluded from this evaluation that the spatial resolution, and SNR are superior to that observed with phosphor screens.

Large-area contrast detectability is similar for the two types of detectors. The experimental results show that the limiting spatial resolution of the SFO system is 8-9 lp/mm compared with 2-6 lp/mm for the phosphor screen system. However, responsivity, or system sensitivity, with the SFO is much lower than measured with phosphor screens. By adjusting the gains of a light amplifier and of an electronic amplifier, an adequately large signal output for a feasible x-ray dose input was obtained with the SFO system. The SFO system has shown images on two kinds of electronic assemblies and met the requirements of the ASTM 801E #4. The experimental results indicate that the SFO detector is comparable to phosphor screens at low energy. An advantage of SFO over traditional phosphor screens are that the trade-off between spatial resolution, SNR and detection efficiency is negligible. At high photon energy, SFO can be made thicker to increase efficiency and SNR, while preserving high spatial resolution. Therefore, SFO is expected to be a more efficient detector and exhibit superior performance for high energy photon detection.

CHAPTER V

ANALYTICAL MODEL AND SIMULATION

5.1 Introduction

This chapter proposes an analytic model which describes the efficiency and spatial resolution properties of SFO detectors. Since the invention of fiber optics, many studies on the transmission properties of cylindrical fibers have been performed [61]. However, the analysis of the light transmission properties of SFO has not been carefully undertaken. SFO, unlike communication optical fibers in which light is admitted only at the fiber ends, typically have a square or hexagonal cross section, and light is generated along the entire active length of scintillating fibers.

As in the model typically used to describe a communication optical fiber waveguides [78], the analytic model in this paper assumes that the light losses in fibers are due to: (1) reflection loss at the end of the SFO, (2) imperfect internal reflections as the light transmitted along the SFO, and (3) light absorption and scattering within the fiber core. The model also assumes that the light energy conversion efficiency, (defined as the fraction of deposited photon energy generated to light), and the light

absorption coefficient of the EMA, ϵ , are constants. These two parameters dominate the performance of a SFO detector. However, very little information is available in the published literature describing these important parameters.

Experiments were performed to measure the detector efficiency and spatial resolution of SFO detectors (Chapter III, IV). By fitting the experimental results to a numerical estimation of the theoretical equations, the absorption coefficient of the EMA and the light conversion efficiency were determined.

The analytical model was used to predict the performance of the scintillating fiber optics for high energy photon detection. The spatial resolution of SFO detector at different detector thickness and different energies were calculated. The simulated results were also compared with the experimental results

The analytical model presented in this chapter was developed for square inorganic scintillating fiber bundles. However, the method is generally applicable and can be used to evaluate any type of scintillating fiber optics.

Selected mechanical and optical properties of the SFO bundles which were used in both the experimental measurements and analytical models, are listed in Table 1.

5.2 Energy Deposition

The energy deposited in the Tb_2O_3 - based SFO was calculated on an IBM-3083 using the ITS code [28]. The ITS code performs a Monte Carlo calculation of the SFO response by taking into account both photon and electron transport. In the analysis it is assumed that the number of incident quanta of radiation is between 10^5 ~ 2×10^6 which ensures that the statistical error in the response is less than 3%. The secondary particles are followed in the SFO until their energies fall below 5% of the incident quanta energy.

The simulation model is shown in Figure 5.1. The gamma-ray impulse penetrates into the front face of the central fiber (#1), with a direction parallel to the fiber axis. As discussed in Chapter II, for gamma-ray interaction with matter in the energy domain of 0.01 to 10 MeV, all but a few minor effects are due to only three of the many competing processes, although the photon-electron cascades are complicated. These are the photoelectric effect, the Compton effect, and pair production. Figure 2.1 provides a handy guide to the relative importance of these three principal interactions over broad ranges of energies of the incident photons and atomic numbers, Z , of the attenuating material. The Tb_2O_3 - based SFO has an effective Z number of approximately 35. The Compton effect dominates for photon energies between 0.9 and 8 MeV. Figures 5.2 and 5.3 show the energy absorption

and scattering calculated by the ITS code in 0.5 cm thick and 2.0 cm thick SDD scintillating fiber optical bundles for impulse photon inputs.

The results in Figures 5.2 and 5.3 indicate that the majority of the incident energy is deposited in the central fiber that is the fiber which received the impulse of radiation. The energy deposited in the fibers contiguous to the central one is mainly due to Compton scattering and pair production, which are the primary factors degradation the spatial resolution of the detector. The simulation results match the analysis in Section 2.6 which means the physical model and the Monte Carlo calculation are consistent.

The energy absorption contribution of Compton scattering and pair production relative to the photoelectric effect increases as the incident photon energy increases. However, as the incident photon energy reaches 4 MeV the area over which incident energy is deposited begins to decrease. This is a consequence of Compton scattering becoming forward peaked [50], and pair production beginning to become the dominant energy deposition mechanism at this energy level. The energy being deposited is more locally in pair production. Noted that Figures 5.2 and 5.3 are similar, indicating that the detector resolution is not strongly dependent on the detector thickness.

Figure 5.1 Simulation Model for Monte Carlo Calculation

Figure 5.2 Energy Deposition and Scattering in 0.5 cm SFO Detector

Figure 5.3 Energy Deposition and Scattering in 2.0 cm SFO Detector

The absorption efficiency, i.e. the fraction of the incident photon energy which has been deposited in the SFO is plotted in Figure 5.4 for 0.5 cm and 2 cm thick SDD scintillating fiber optical detectors. The percentage of the energy deposited in the SFO is a design parameters of the SFO. Figure 5.4 shows the minimum absorption of the SFO occurs at approximately 5 MeV. This occurs because the scintillating material has the lowest interaction cross section at this energy.

Figure 5.4 Absorption Efficiency of SFO

5.3 Detector Efficiency

Analysis of the detector efficiency is based on ray trace methods for the scintillating light which incorporate complete internal reflection at the core/cladding interface. The energy carried by the cladding is not taken into account. This is a good approximation for SFO because most of the light is transmitted in the fiber core since SFO's are high multimode fibers [9]

To analyze SFO detector efficiency it is first necessary to analyze how the light is guided through a square waveguide. Assume an arbitrary ray, as shown in Figure 5.5, is generated inside a fiber at point G and the direction of propagation is expressed by

$$\hat{r} = \sin \theta \cos \varphi \hat{x} + \sin \theta \sin \varphi \hat{y} + \cos \theta \hat{z}, \quad (5.1)$$

where θ is the angle between the direction of propagation and z-axis, and φ is the angle between the projection of propagation onto x-y plane and x-axis as shown in Figure 5.5. The ray representing the direction of propagation intersects the fiber core/cladding interface at point P on the y-z plane, where it can be expressed as

$$\hat{r} = \cos \psi \hat{x} + \sin \psi \cos \gamma \hat{y} + \sin \psi \sin \gamma \hat{z}. \quad (5.2)$$

ψ is the angle between the direction of light propagation and the surface normal, i.e., the incident angle, and γ is the angle between the y-z plane projection of the ray and z-axis. Comparing Equations 5.1 and 5.2,

$$\cos \psi = \sin \theta \cos \varphi. \quad (5.3)$$

The total reflection at P requires

$$n_{\text{core}} \sin \psi \geq n_{\text{clad}}, \quad (5.4)$$

Therefore,

$$\cos \psi = \sin \theta \cos \varphi \leq \sqrt{1 - (n_{\text{core}} / n_{\text{clad}})^2}. \quad (5.5)$$

Equation 5.5 holds for any ray, skew or meridional, at any fiber core/cladding interface in a square fiber. A ray with an incident angle ψ which obeys Equation 5.5 will be reflected back into the fiber with angle of reflection equal to angle of incidence. The maximum angle for which a ray can be reflected is

$$\cos \psi_c = \sin \theta \cos \varphi = \sqrt{1 - (n_{\text{clad}} / n_{\text{core}})^2}. \quad (5.6)$$

For a meridional treatment in which skew light is not considered, $\cos \varphi = 1$ and Equation 5.6 is simply

$$\cos \psi_c = \sin \theta_c = \sqrt{1 - (n_{\text{clad}} / n_{\text{core}})^2}, \quad (5.7)$$

where θ_c is the critical angle for total reflection. These results are the same as those developed for round fibers [61].

Figure 5.5 Total Reflection of A Ray in A Square Waveguide

Scintillating fiber optics used as image detectors convert gamma or x-ray photons into light photons which can be more easily transmitted and detected. The interaction of the gamma or x-ray photons with the SFO generate charged particles-electrons through the photoelectric effect, the Compton effect, and pair production. During the slowing down of these electrons, some of the atoms (or molecules) of the scintillators are excited; that is, their electrons are transferred from the ground state into excited states. Scintillating light is then generated following deexcitation. The exact nature of this process is quite complex [56], therefore, the

scintillating light is usually assumed to be generated locally and isotropically, but not instantaneously. The decay time of the excited state usually determines the response time.

As illustrated in Figure 5.6, a scintillating fiber of length L is irradiated by a parallel photon beam and the interaction occurring at a distance d from the incident face of the detector. Assuming an incoming photon has energy I_0 and the light energy conversion efficiency is η_c , then the scintillating light generated at the distance d is $\eta_c I_0$.

Figure 5.6 Light Generating and Transmission in SFO

The scintillating photons with small enough deviations from the fiber direction will be guided to both ends of the fiber, reflecting at the core/cladding interface with a constant reflection angle. Assuming α is the light attenuation coefficient for the fiber core in the z direction, and a ray with an angle θ with the z axis travels a distance $1/\cos\theta$ through the fiber core per unit length of fiber, thus the absorption loss between the scintillating position d and the fiber end location L is

$$e^{-\frac{\alpha(L-d)}{\cos\theta}}. \quad (5.8)$$

If β is the reflection coefficient of the core/cladding interface, and the number of reflections, n, in unit length of the fiber is

$$n = \frac{\tan\theta}{D \cdot \cos\theta}, \quad (5.9)$$

where, D is the diameter of the fiber. The reflection loss after the light traveling of a distance of L-d is,

$$\beta^{n(L-d)} = \beta^{\frac{\tan\theta}{D \cos\theta}(L-d)}. \quad (5.10)$$

A formal expression for the total flux transmitted by a straight circular cylindrical fiber can be developed by integrating over the

angular distribution of the light and the area of a fiber which can conduct the light [61],

$$F = \iint tI(\theta, \varphi) dA d\Omega. \quad (5.11)$$

In Equation 5.11, F is the total flux of light, t is the transmission function which depends on the properties of the fiber and the orientation of the ray, $I(\theta, \varphi)$ is the angular distribution of the light in the medium of the fiber core, θ and φ have their usual meaning in spherical coordinates, dA is the element of area in the cross section, and $d\Omega$ is the element of solid angle. This formulation assumes that all rays are incoherent and there is no possibility of any kind of interference. The integral is over the intensity of each ray, and no phase information is considered.

Equation 5.11 can be applied to the scintillation fiber optics, where the source of light to be transmitted is internally produced by a radiation process. Assuming isotropic scintillation at d yields

$$I(\theta, \varphi, d) = I(d) = \frac{1}{4 \cdot \pi} \eta_c \cdot I_0. \quad (5.12)$$

As shown in Figure 5.6, half of the guided light travels toward the end of the fiber with the detector. The fiber end is coupled directly to a photomultiplier tube (PMT) with index matching grease. Therefore, negligible reflection loss occurs at the fiber end. The other half of the guided light travels to the opposite end of the fiber, and is reflected back

with reflection loss. The total distance the reflected light travels along z axis is $d+L$. Then, the light intensity at the detector end of the fiber due to the scintillation at d is

$$I(d) = \frac{1}{\pi} I_0 \eta_c \int_0^{\theta_c} \int_0^{2\pi} \left\{ e^{-\frac{\alpha(L-d)}{\cos\theta} \beta \frac{\tan\theta}{D \cos\phi} (L-d)} + (1-T) e^{-\frac{\alpha(L+d)}{\cos\theta} \beta \frac{\tan\theta}{D \cos\phi} (L+d)} \right\} \sin\theta d\theta d\phi, \quad (5.13)$$

where $1-T$ is the reflection loss at the front end of the SFO [79]. If this end is covered by a reflector, the loss term is approximately zero and the value of the output will be nearly doubled. As shown in Equation 5.7, the maximum angle of deviation for which the light will be guided in a square fiber is the same as that in a round fiber; hence, polar coordinates are used to simplify the integration in Equation 5.13.

To find the light resulting from a large number of gamma-ray photons impinging on the detector, the probability density function, $p(z)$ is needed. $p(z)$ is the density function of the point at which the gamma-ray photons interact and expressed as

$$p(z) = \mu e^{-\mu z}. \quad (5.14)$$

where μ is the linear attenuation coefficient of the SFO material.

Integrating over $p(z)$ gives the total output of light

$$I = \int_0^L p(z) I(z) dz. \quad (5.15)$$

For an x-ray input, which is generated, for example, by x-ray tubes, Equation 5.15 becomes

$$I = \int_{E_1}^{E_2} \int_0^L S(E) \mu(E) I(z) e^{-\mu(E)z} dz dE, \quad (5.16)$$

where, $S(E)$ is the x-ray energy spectrum.

The absorption coefficient α is strongly wavelength dependent, and β is slightly wavelength dependent [7]. α and β used here are the average values. If wavelength effect is considered,

$$I = \int_{E_1}^{E_2} \int_0^L \int_{\lambda_1}^{\lambda_2} S(E) \mu(E) I(z, \lambda) e^{-\mu(E)z} dz dE d\lambda. \quad (5.17)$$

Usually, the values of α and β are unknown, but the total attenuation length, that is, the combined effect of both absorption and reflection can be experimentally determined. The attenuation and reflection term can be rearranged to obtain the total attenuation term,

$$e^{-\frac{\alpha}{\cos\theta}\beta\frac{\tan\theta}{d}} = e^{-\frac{\alpha}{\cos\theta}} e^{-\ell n\beta\frac{\tan\theta}{D\cos\varphi}} = e^{-\Lambda}, \quad (5.18)$$

Λ is the total attenuation per unit length which accounts for both attenuation and reflection losses, and θ is the average reflection angle, and $\cos\varphi$ is assumed to be unit.

Because SFO fibers usually have a square or hexagonal cross section, the interstitial area between the fibers can be ignored. Therefore the transmittance of a bundle and a single fiber are almost identical. However, the fraction of core area in the SFO is not considered in the previous analysis, therefore, η_c must be considered as the light energy conversion efficiency of the total SFO bundle (core, cladding, and EMA) rather than the scintillating core material.

The detector efficiency of SFO was experimentally measured [9] and defined by

$$DE = I_{\text{out}} / I_{\text{in}} = \eta_{\alpha}\eta_c\eta_t. \quad (5.19)$$

The absorption coefficient η_{α} was determined in Section 2.1. By using Equation 5.16, the light conversion efficiency η_c , and light transmission efficiency η_t can be determined. The calculated results based on the measured result from the Am^{241} source are listed in Table 5, which is more accurate than obtained in Table 3 in Chapter III.

Table 5 Efficiency Calculation of SFO (%)

	LG9	SDD
light conversion efficiency η_c	7.6	6.4
light transmission efficiency η_t^*	3.8	2.4
light transmission efficiency η_t^{**}	6.4	4.6

* without reflector at the entrance window;

** with reflector at the entrance window.

After η_c is determined, the relationship between the detector efficiency, detector thickness and incident photon energy can be found analytically by Equation 5.16, which is important for the detector design. Figure 5.7 shows the detector efficiency of LG9 for a 1 MeV photon source. When the detector thickness equals 5 cm, the detector efficiency reaches a maximum. Because the model assume η_c is a constant, the curve in Figure 5.7 implicates the relation of attenuation loss with detector thickness, that is, the trade off between transmission efficiency and absorption efficiency. Figure 5.8 is the detector efficiency of SFO at different incident photon energies. The detector efficiency will decrease with an increase in photon energy. However, the detector efficiency will increase slightly as the energy exceeds 5 MeV because the photon cross section of the SFO begins to increase as discussed previously in Section 5.2.

Figure 5.7 The Efficiency of LG9 for 1 MeV Gamma-ray Detection

Figure 5.8 The Efficiency of LG9 for Different Photon Energy Detection

5.4 Spatial Resolution

The spatial resolution of SFO depends on energy scattering, fiber diameter and cross talk between fibers. The scattering effect has discussed in Section 5.2.

At low photon energies, the photoelectric effect is dominant and the effect of secondary scattering interaction in the scintillating detector can be neglected. However, at high photon energies, the dominant effect is the

Compton effect, therefore, the secondary scattering has to be included. In this section, first, a model for SFO photon detection at low energy is developed. The model is then extended to high energy photon detection. Spatial resolution analysis for phosphor screens are also developed for comparison.

The SFO bundle is simplified as shown in Figure 5.9. The cladding is ignored and the square fiber is simplified to a round circles. Because the unsharpness of SFO was determined experimentally to be on the order of $100\ \mu\text{m}$ and the size of the fiber is less than $20\ \mu\text{m}$ as shown in Chapter III [80]; therefore, the shape of the fibers is not important, and this simplification is reasonable.

Figure 5.9 The Physical Model of SFO for Spatial Resolution Analysis

Figure 5.10 X-ray Scintillating Process of SFO

The simulated model is shown as Figure 5.10. The impulse input is a monoenergetic parallel photon beam which impinges at the center of the SFO and generates the scintillation at a distance d from the SFO front face. Assuming the generated light propagates uniformly in the SFO, and the granular nature of the phosphor is ignored, the guided light with a angle smaller than the critical angle will be guided through the fiber as discussed in Section 2.2. However, if the light is unguided, the light will penetrate to other fibers through the EMA. As the stray light passes through a layer of the EMA, the intensity will be reduced by ϵ , the

absorption coefficient of the EMA, although the direction of the ray remains unchanged.

It is intuitively obvious that the fiber bundle is not capable of giving a faithful image of an object which has a spatial period smaller than the diameter of fibers. Therefore, there is little reason, when considering fiber bundles, to be concerned about the exact shape of the point spread function in a single fiber. The multiple reflections occurring within an individual fiber tend to integrate any variations in illumination across its entrance aperture so that the exit aperture of the fiber will appear uniformly bright.

As shown in Equation 5.7 , the critical angle for the internal reflection is

$$\theta_c = \cos^{-1}(n_{\text{clad}} / n_{\text{core}}). \quad (5.20)$$

If the scintillating light ray with an angle,

$$\theta = \tan^{-1}\left(\frac{r}{L-d}\right) \leq \theta_c, \quad (5.21)$$

the light is guided. For the guided light, the fiber core guides the light with attenuation due to absorption and reflection losses as shown in Equations 5.8 and 5.10. The point response function (PSF) which is contributed by the guided light generated at point d is

$$h_0(r, d) = h(r = 0, d) = \frac{1}{2} \int_0^{\theta_c} e^{-\frac{\alpha(L-d)}{\cos\theta}} \frac{\tan\theta}{\beta D^{(L-d)}} \sin\theta d\theta, \quad (5.22)$$

where h_0 means the PSF at the center, and $1/2$ means only the light guided to the right end of the fiber is counted. The integral includes the attenuation losses and integrates from angle $\theta = 0$ up to the critical angle θ_c .

For $r \neq 0$, the corresponding point response functions $h_1(r, z), h_2(r, z), h_3(r, z), \dots$ are chosen at points outside the central as shown in Figure 5.10, and are contributed by the unguided light. These point response functions along the light detector are governed by two factors, the obliquity factor of the impinging light photons, which follows a cosine law, and the inverse-square-law falloff with distance from the scintillating point, which has a $\cos^2 \theta$ dependence [40]. Thus, the PSF at the point of the i is

$$h_i(r, d) = \frac{1}{4\pi} \cos^3 \theta e^{-\alpha(L-d)/\cos\theta} \epsilon^i, \quad (5.23)$$

where i also means the unguided light has passed i layers of EMA, $\frac{1}{4\pi}$ means the isotropical scintillating light, and the exponential term represents the absorption loss. Note the reflection loss has no effect for the unguided light and, therefore, has been ignored. Because

$$\cos \theta = \frac{L - d}{\sqrt{(L - d)^2 + r^2}}, \quad (5.24)$$

therefore,

$$h_i(r, d) = \frac{1}{4\pi} \frac{(L - d)^3}{[(L - d)^2 + d^2]^{3/2}} e^{-\alpha[(L-d)^2 + r^2]^{1/2}} \epsilon^i. \quad (5.25)$$

To find the average PSF $h_i(r)$ resulting from scintillation from $z=0$ to L by a large number of x-ray photons, integrate over the probability density function $p(z)$,

$$h_i(r) = \int_0^L p(z) h_i(r, z) dz = \mu \int_0^L h_i(r, z) e^{-\mu z} dz. \quad (5.26)$$

For the x-ray input, the average PSF can be expressed as

$$h_i(r) = \int_{E_1}^{E_2} \int_0^L S(E) \mu(E) h_i(r, z, E) e^{-\mu(E)z} dz dE, \quad (5.27)$$

It is often favorable to obtain the frequency response of the image detector. Since the system is linear and spatially invariant, the frequency response of the system is given by the Fourier transform of the point

spread function. The Fourier transfer function in polar coordinates is given by Hankel transform. In the case of circular symmetry the resultant transfer function becomes

$$G(\rho, \theta) = G(\rho) = \int_0^{\infty} g_R(r) J_0(2\pi r \rho) 2\pi r dr. \quad (5.28)$$

where, $J_0(2\pi r \rho)$ is the Kernel of the Fourier-Bessel transform for the functions having circular symmetry and ρ is the radial spatial frequency variable. The Hankel transform of zero order is often referred to as the Fourier-Bessel transform. Therefore, the frequency response of the system is

$$\begin{aligned} H(\rho) &= \int_0^{\infty} h(r) J_0(2\pi r \rho) 2\pi r dr \\ &= \left[\int_0^{R_1} h_0(r) + \int_{R_1}^{R_2} h_1(r) + \dots \right] J_0(2\pi r \rho) 2\pi r dr, \end{aligned} \quad (5.29)$$

It is convenient to use a normalized frequency response $\bar{H}(\rho)$ as given by

$$\bar{H}(\rho) = H(\rho) / H(0). \quad (5.30)$$

For high energy photon detection where the scattering has to be considered, the results from Section 5.2 can be used. The energies deposited in the fibers around the center one is simulated by the ITS code as shown in Figures 5.2 and 5.3. The point spread function at point i is considered to be the contribution of the distributed energies, that is, the weighed sum of point spread functions resulting from the energies deposited at different fibers. The quantum of these energies deposited can be considered as weighting factors. Therefore, Equation 5.26 can be modified

$$h_i(r) = \sum_j k_j h_j(r - i\Delta r), \quad (5.31)$$

where $h_j(r - i\Delta r)$ is the average PSF of i th fiber next to the center one, and k_i is the weighting factor.

The physical model of phosphor screens is illustrated in Figure 5.11. The point response function is

$$\begin{aligned} h(r, z) &= h(0, z) \cos^3 \theta e^{-\alpha(L-z)/\cos \theta} \\ &= \frac{1}{4\pi} \frac{(L-z)^3}{[(L-z)^2 + r^2]^{3/2}} e^{-\alpha[(L-z)^2 + r^2]^{1/2}}, \end{aligned} \quad (5.32)$$

and the average PSF is

$$\begin{aligned}
 h(r) &= \int_0^L p(z) h(r, z) dz \\
 &= \frac{1}{4\pi} \mu \int_0^L \frac{(L-z)^3}{[(L-z)^2 + r^2]^{3/2}} e^{-\alpha[(L-z)^2 + r^2]^{1/2}} e^{-\mu z} dz. \quad (5.33)
 \end{aligned}$$

Figure 5.11 X-ray Scintillating Process of Phosphor Screens

Figure 5.12 shows the modulated transfer functions (MTF) of scintillating fiber optics from the experimental and calculated results. The experimental MTF was calculated digitally from the Fourier transform of the measured edge response function. The ERFs were sampled at a discrete set of points. The difference between the resulting

MTF and the true MTF is due to the distance between ESF samples. The curves in Figure 5.12 is the MTF of LG9 scintillating fiber optics generated by 90 keV x-ray. The theoretical curve was calculated by Equation 5.30 in which the absorption coefficient of the EMA is assumed to be 0.05. The experimental result compares well with the calculated curve. Similar results can be obtained for SDD detectors. The absorption coefficient of the SDD's EMA is also assumed to be 0.05.

The EMA plays a dominant role in determining the spatial resolution of SFO as shown in Figures 5.13 and 5.14. When the absorption coefficient of EMA changes from 0.0 to 0.1, the limiting spatial resolution changes from 2 lp/mm to 15 lp/mm. Therefore, it is important to increase the absorption coefficient in EMA for high spatial resolution.

Simulation detector spatial resolution at different energies and at different detector thickness are shown in Figures 5.15 and 5.16. Figure 5.15 is the point spread function and Figure 5.16 is the respective MTF. The results show that the spatial resolution has no significant decrease, when

- (1). the detector thickness increases from 0.5 cm to 2.0 cm,
- (2). or/and the photon energy increases from 0.1 MeV to 1.0 MeV.

The first evidence was experimentally determined in Chapter III. Figure 3.9 shows that the spatial resolution of 0.5 cm and 1.0 cm thickness SDD detectors are almost identical and can not be distinguished due to the instrumental resolution and statistic noise. This thickness independent phenomena is primarily because of the absorption of unguided photons by

the EMA. The unguided light will be reduced 5% as it passes an fiber. Therefore, when the light passes a few millimeters in the detector medium, the light will be eliminated and an increase in thickness from 5 mm to 20 mm results in no degradation in spatial resolution.

The edge response function for 6 and 18 MeV x-rays were measured by recording the spatial distribution of the light on films. The unsharpness was estimated to be 340 μm . That is approximated double the low energy unsharpness of 170 μm . The simulated results indicate no significant reduction in spatial resolution with increasing the photon energy. The measured degradation in spatial resolution for 6 and 18 MeV is a consequence fo photon backscattering resulting from the experimental setup and can be eliminated in a imaging configuration.

Figure 5.12 The MTF of LG9 SFO Detector at 90 keV

Figure 5.13 The LSF of 0.5 cm SDD Detector at 90 keV

Figure 5.14 The MTF of 0.5 cm SDD Detector

(a) 0.5 cm SDD

(b) 2.0 cm SDD

Figure 5.15 Point Spread Functions of SDD Detectors for 0.1 and 1 MeV photons

Figure 5.16 The MTF of SDD Detectors for 0.1 and 1 MeV photons

5.5 Conclusions

The light transmission properties of SFO have been analyzed. The energy absorption and scattering in SFO is first simulated by using the ITS Monte Carlo code. The dependence of absorption and scattering on photon energy and detector thickness are demonstrated. To find the analytical model of SFO, an analysis of the light generation and

transmission is developed based on the ray method. Then, the equations which describe the light transmission and detector efficiency are derived.

A series of experiments were performed on two types of SFO to measure the detector efficiency and spatial resolution. By fitting the experimental data to the analytical results, the light conversion efficiency and the absorption coefficient of EMA were found. After the light conversion efficiency was found, the detector efficiency at different energies was calculated and plotted. The simulated results indicate that the spatial resolution will not be significantly reduced with either increase photon energy or increase detector thickness. This conclusion has been compared with the experiment result in Chapter III in which an experiment using 6 and 18 MeV x-rays. The analysis also shows that the EMA is a very important factor in determining the spatial resolution of the SFO detector.

CHAPTER VI

QUANTITATIVE METHODS FOR X-RAY IMAGING

6.1 Introduction

This chapter describes quantitative methods that determine the ability of a SFO detector array in detecting a low contrast lesion from a uniform background for both low and high energy x-ray imaging.

To judge the best diagnostic image quality means to obtain the best final output of a system from a given input. The quantitative methods are various; however, each should be useful for assessing and comparing the image quality, predicting the limitation of a given system, and for optimizing design. The parameters described in Chapter V might be called the inherent image quality of the image system. However, they do not relate directly to their diagnostic image quality. For example, the MTF is a measure of certain optical characteristics of an imaging system. However, it provides no information regarding the effect of noise and radiographic contrast on the image [81]. Since both of these phenomena affect the visibility of detail and since the quantum noise depends on the MTF, excellent modulation transfer characteristics, which can be

considered an attribute of excellent inherent image quality, do not necessarily result in excellent visibility of detail and in excellent diagnostic image quality [82]. Measurement of MTF alone may result in erroneous conclusions regarding the diagnostic image quality of the system. Similarly, any set of system parameters, no matter how complete, measured by purely physical means may not give a correct indication of diagnostic image quality.

The entire image quality of an image system usually depends not only on the image system, but also the entire radiological process. In general, the entire radiological process involves:

- (1) exposing operation, determined by the quality of the x-ray or the gamma ray beams, absorption and scattering properties of the object, and the geometry of the system setup;
- (2) recording system, determined by the transfer functions of the image recording system, such as those of the image detector, the optical and electrical intensifier;
- (3) visual and recognition system, determined by the observer's eyes, illuminator or other optical display system, and by the experience and pattern recognition ability of the radiologist.

One of the objectives of our research is to evaluate the image capability of the SFO image detector. The SFO detector is part of the recording system. The effects from additional instrumentation and subjective judgment, such as exposing operation and image intensifiers, and the recognition by human eyes and experience, should be avoided. In addition, quantitative methodology was developed originally from the

conventional radiographic system in which the film density is a primary factor. The method we choose must be suitable to the digital system in which the image is displayed by an optical-electron display monitor but not a film. There are different quantitative methodologies with their advantages and disadvantages. They will be introduced and compared in the next section.

6.2 Quantitative Methods Comparison

In traditional x-ray imaging systems, the actual spatial distributions of images are displayed on a cathode ray tube (CRT) and recorded on photographic films. Thus, contrast in the distribution of counts presented to the display is translated into an optical density contrast by the film. The first method is based on image contrast and visual perception of this contrast in photographic film. Assume the optical density difference, ΔD , between two equal areas is given by [83]

$$\Delta D = D_l - D_b = \chi \log(N_l / N_b), \quad (6.1)$$

in which D_l and D_b are the optical density of the film, N_l and N_b are the count densities corresponding to the lesion and background, and χ is a film dependent parameter, typically 1.35 to 2.0. The optical density difference is visualized as a contrast difference in the film when the contrast level exceeds the threshold level for human perception.

The perception of a low contrast lesion depends on the threshold of human vision and the contrast available in the film. The visual interpretation of the image is limited by the human eye-brain response, the characteristics of photographic film, and the conventional methods of display [84]. Therefore, this method is limited by human decision and film threshold level.

The second method called statistical image interpretation is based on statistical signal detection theory and describes the effects of a finite number of image counts on the detectability of low contrast lesions [85]. Signal detection theory was originally developed for the analysis of radar and information system. The application in radiography involves determining the presence of a single lesion with a given size, location, and contrast. If an ideal statistical observer is assumed, the observer's response is strictly determined by the statistical information that is available. The probability of detection is a function of contrast, spatial resolution, and statistics. For a given image system, the detectability of a minimum size and contrast is then determined.

The quality of an image system can also be studied by the output spatial frequency spectrum of the image system [87]. The input to radiographic system must be a known x-ray pattern. The x-ray pattern is described by the spatial distribution of photon fluence and the spatial composition of x-ray radiation. By applying concepts of MTF, LSF, and noise spectrum, image parameters relating the system output to the input x-ray pattern can be found. A further advantage of this method is that it related to the noise characteristics (Wiener spectrum) of imaging systems.

The quantitative information of this study is from the analysis of experimental data. However, unless the x-ray patterns can reflect the diagnostically important detail, the information is not of much value.

In contrast to the spatial frequency spectrum method, evaluation of the system quality by the resolving power function is a pure theoretical approach [88]. The resolving power function expresses the ability of the system to detect changes in film density that correspond to changes in the property of the examined item. The resolving power function is a characteristic function of the radiographic system, and depends on the various parameters of the system. Thus, it reflects the influence of these parameters on the image. The method presented is a useful tool for predicting the limit of detection, contrast reduction and resolution of a practical system. However, the derivation of the resolving power function is based on the film density. If applied to digital radiographic system, the resolving power function has to be redefined.

The figure of merit (FOM) method is based on the two major physical properties of an imaging system, the MTF and the signal to noise ratio. The concept of this evaluation criteria was first developed by A. Ginsburg [89]. Then, the method was applied by R. Polichar [3] to provide an indication in detection defects in a continues background. The detail of this method is further introduced in the next section.

The figure of merit method is used in evaluation of the SFO image detectors for the following reasons:

- (1) the method is based on the basic inherent image quality, MTF and SNR, which can be uniquely defined and measured;

- (2) the method is based on experimental data , and no subjective judgment is involved;
- (3) the method is not based on film radiography, and as analyzed in the next section, reflects on the characteristics of a digital image system;
- (4) the performance of SFO at any photon energy level can be predicted by simulation of the combined behavior of the MTF and SNR;
- (5) The quantitative information characterizes the image quality on the measurement of dimensions of details, which reflects the image capability for detecting defects of different sizes.

However, this methodology does have some drawbacks. For example, it only reflects the effect of the MTF and SNR. Other parameters, such as the detector efficiency, which may affect the image quality, are not reflected on the evaluation. As mentioned in Chapter IV, large signal output was obtained by using a combination of a SFO detector and a light amplifier. Therefore, the efficiency requirement is not a critical factor for x-ray imaging.

6.3 Figure of Merit

The figure of merit was proposed by R. Polichar [3] as

$$\text{FOM}(D, E) = D / p \cdot \text{SNR}(E) \cdot S(\text{MTF}(E), D), \quad (6.2)$$

where, p is the linear dimension of the pixel, and D is the diameter of a small spatial detail (defect) presented by a single disk. D/p is

proportional to the square root of the number of the pixels in the disk. The smearing factor $S(\text{MTF}(E), D)$ is the overlap integral that determines the effect of $\text{MTF}(E)$ on the disk with a diameter D . The smearing factor will be described in more detail in the later paragraph. Because both the spatial resolution and signal to noise ratio are energy dependent, the figure of merit usually changes with the energy E .

As shown in Equation 6.2, the figure of merit combines the smearing effect of the MTF and the noise structure information to provide an information on system imaging capability. This model is based on system's physical and statistical principles and is accurate only for small defect detection.

The ability of an observer to detect an abnormality in an image depends both on the signal to noise ratio associated with that abnormality and on the modulation transfer function [3]. Consider an image with uniform background intensity density of $\bar{h}$ counts per unit, the count density will be changed to $\bar{h} \pm \Delta h$ when contains a disk. For small disk, the threshold contrast $\Delta h / \bar{h}$ is proportional to the signal to noise ratio for poisson statistics. For high noise and low contrast imaging, the threshold contrast determines whether or not the small disk image is being detected. Also, the threshold contrast is not related to the system MTF which only implies the system ability in detecting fine details . On the other hand, when the disk is small, the MTF has a significant effect on the imaging quality. This effect is presented by the smearing factor $S(\text{MTF}(E), D)$, and can be separated from SNR which affects the threshold contrast. In this case, MTF and SNR are assumed to be uncorrelated. Therefore, the

change of the signal noise ratio does not affect the MTF, and vice versa.

Suppose there are two different image systems, and each characterized by their MTF and SNR. The method predicts which system has a better image quality in terms of distinguishing small spatial details by comparing from their figure of merit curves.

The smearing factor $S(\text{MTF}(E), D)$ is developed as follows. The disk can be defined as a radially symmetric rectangular function,

$$F_D(r) = \begin{cases} 1 & \text{for } r < D/2 \\ 0 & \text{for } r > D/2 \end{cases} \quad (6.3)$$

The effect of the $\text{MTF}(E)$ on the rectangular function $F_D(r)$ results in the distribution of the output intensity. Usually if the system is radially symmetric, the distribution is also radially symmetric. The intensity distribution function in space can be calculated according the image theory developed in Section 2.3. For a 1-D x-ray image, the output distribution over the disk $F_D(x)$ can be obtained from the 1-D convolution integral as

$$I(x) = F_D(x) * \text{LSF}(x) = \int_{-\infty}^{+\infty} F_D(\xi) \text{LSF}(x - \xi) d\xi, \quad (6.4)$$

where the $\text{LSF}(x)$ is the line spread function of the detector. The distribution can also be calculated by using the spatial frequency domain,

$$\begin{aligned}
\tilde{I}(x) &= F^{-1}F[F_D(x) * LSF(x)] \\
&= F^{-1}\{F[F_D(x)]MTF(E)\} \\
&= F^{-1}\left\{D \frac{\sin \pi Df}{\pi Df} MTF(E)\right\},
\end{aligned} \tag{6.5}$$

in which $D \frac{\sin \pi Df}{\pi Df}$ is the Fourier transform of the function $F_D(x)$. The factor $S(MTF, D)$ is the integral of the distribution function,

$$S(MTF, D) = \int_{-\infty}^{+\infty} I(x) dx \tag{6.6}$$

As analyzed in Chapter IV, the scintillating detector is the dominant factor in determining the SNR of a SFO based on a DADR system. For this simulation, the SNR measured from the DADR system is assumed to be the detector's SNR. Because all the data are obtained from the same image system, p is a constant and a relative FOM can be applied to simplify the calculation,

$$FOM_T(D, E) = D \cdot SNR(E) \cdot S(MTF(E), D). \tag{6.7}$$

In this case, the figure of merit for a specified defect diameter is determined only by the signal to noise ratio and the MTF.

The calculation of FOM was implemented on a Sun workstation by using the MATLAB package [95]. The signal to noise ratio is evaluated at 90 keV by using a 8 mm aluminum absorber. As shown in Figure 6.1, the figure of merits of SFO is overall higher than those of phosphor screens. This is a consequence of both the MTF and SNR for the SFO being better than observed in the phosphor screens. The FOM for the SDD is slightly higher than that of the LG9. This is because SDD is made of denser materials which results in a higher SNR while their spatial resolutions are very similar. However, better FOM from the LG9 can be obtained by increasing the thickness of LG9. In this case a better SNR is obtained without losing much spatial resolution. From a close look of Figure 6.1, the two FOM curves of SDD and LG9 are closer in the smaller defect diameter region than in the larger diameter region because LG9 has better spatial resolution that compensates for its lower SNR and results in a better FOM for small defect detection.

Figure 6.1 The Figure of Merits for Different Image Detectors
for 90 keV X-rays

6.4 The FOM at High Photon Energies

The model developed in the previous section describes both the effects of the MTF and SNR on the image quality for varying size defects. In this section, the model is extended to simulate the behavior of scintillating fiber optical detectors for high energy photon detection.

The MTF of scintillating fiber optical detector for high photon energy was simulated in Chapter V, and includes scattering effects. The signal to noise ratio for high photon energy level, however, can not be easily measured or simulated. As discussed in Chapter IV, the signal-to-noise of a system with q successive stages can be expressed as

$$\text{SNR} \propto \frac{\sqrt{\eta_a N}}{\sqrt{1 + \sum_{i=1}^q 1 / \prod_{j=1}^i g_j}}. \quad (6.8)$$

General speaking, the product term, $g_1 g_2 g_3, \dots$, is appreciably greater than unity, and the number of photons per pixel element impinging on the detector, N , is identical for the same system operated at same conditions. Therefore, the approximate signal-to-noise ratio is,

$$\text{SNR} \propto \sqrt{\eta_a}. \quad (6.9)$$

The SNR is proportional to the square root of the detector absorption efficiency, η_a . This is also an energy dependent parameter. The equation for the relative figure of merit is then replaced by

$$\text{FOM}_r(D, E) = D \sqrt{\eta_a(E)} \cdot S(\text{MTF}(E), D). \quad (6.10)$$

SDD detectors of thickness 0.5 cm and 2.0 cm are simulated and the results are shown in Figure 6.2. The curves show clearly the energy effect on the capability of image detectors. The decrease of FOM at high energy is due to the decrease in the interaction rate (absorption efficiency). However, the interaction rate can be increased by increasing the thickness of the SFO detector so that FOM is increased as shown in Figure 6.2. As discussed in Chapter V, the spatial resolution will not be significantly affected by changing the photon energy and detector thickness. Therefore, the FOM is improved by increasing the detector thickness from 5 mm to 20 mm for both 1 MeV and 5 MeV photon imaging. On the other hand, for 100 keV photon imaging the FOM is reduced as the detector thickness increased. This is the consequence of the slight decrease in spatial resolution and no increase in SNR by increasing the detector thickness at this photon energy level. In principle, the detector thickness can be increased so that η_a is always close to unity for any energy detection. In practice, other factors, such as the cost and the self-absorption of the thicker detectors, have to be considered.

This simulation is a predictor of SFO performance for high energy photon detection. It shows that the imaging capability of the SFO

detectors can be improved by adjusting their thickness. Phosphor screens, on the other hand, can not make this adjustment because the trade off between spatial resolution and SNR. The design of a SFO for a specified photon energy can be optimized by maximizing their FOMs for different detector thicknesses.

Figure 6.2 The Figure of Merits of SDD at Different Photon Energies

6.5 Summary

The objective of this chapter was to evaluate the imaging capability of SFO image detectors by using quantitative methods. Different quantitative methods were first presented and compared. The figure of merit method that combines the detector MTF and SNR was chosen for this research. The main advantage of the FOM method is that inherent unknown factors that affect the evaluation result, such as subject judgment and instruments, are removed. The results obtained in Chapters III, IV and V were directly used in this simulation. A comparison of SFO detectors with conventional screens shows that the FOM for SFO detectors is overall better than for phosphor screens.

The FOM method was then extended to investigate the imaging capabilities of the SFO detector for high energy photon. The simulation shows the imaging capability of the SFO detector can be improved by increasing the detector thickness. The FOM method can be used to design the image detector by optimizing the detector thickness.

CHAPTER VII

SUMMARY AND CONCLUSIONS

7.1 Summary

The overall goal of this dissertation, to evaluate the scintillating fiber optics and their applications in radiographic systems has been met and demonstrated.

Chapter I proposed the objective of this research, along with a review of the development of scintillating fiber optics and a preview of the following chapters.

The general theory was described in Chapter II. Emphasis of this discussion is directed to the basic concept of scintillating fiber optics.

The experimental evaluation of SFO detectors is described in Chapter III. The basic properties of Tb₂O₃ scintillating fiber optics were also investigated.

In Chapter IV, the application of SFO image detectors in the Scantech DADR system is presented. The overall performance of the SFO compared with phosphor screens was evaluated in the DADR real time digital radiographic system.

A computer simulation of the SFO detector was developed in Chapter V. A combination of Monte Carlo calculations and an analytic model simulates the detector efficiency and spatial resolution. Secondary effects such as scattered radiation were also considered in the modeling.

In Chapter VI, the figure of merits quantitative method of analysis based on the use of a computer simulation was developed to investigate the imaging capabilities of SFO image detectors. The main advantage of the figure of merit method is that the inherent unknowns of image procedures and subject judgments are removed from the investigation.

7.2 Conclusion

The difficulty for high energy photon detection is that as the photon energy increases, the contrast between areas of differing attenuation decreases. As contrast decreases, the number of photon interactions necessary to resolve the contrast increases. This conflicts with the fact that the detector efficiency decreases with increased photon energy. Therefore, more interactions are required with a less efficient detector as the photon energy increases. To solve this problem, a thicker detector has to be used. However, conventional screen detectors result in degradation of spatial resolution due to the spreading of the scintillating light.

The advantage of the scintillating fiber optics is that the detector thickness can be increased to increase high interaction rate and detection efficiency while retaining spatial resolution. This is because scintillating

fiber optics combine scintillating material with a dielectric waveguide. This configuration permits an increase in thickness of the scintillating material without significant dispersion of light, therefore preserving spatial resolution with increased interaction rate and detection efficiency. This characteristic is advantageous for imaging of high energy photons. It must be noted that this advantage is based on the tradeoff in loss of detector light transmission efficiency.

Two types of terbium oxide glass scintillating fiber optics were evaluated. Besides the waveguide function of the optical fibers, the scintillating fiber optics exhibits the properties of ordinary terbium oxide scintillating glass. The absorption efficiency of SFO can be made close to unity when high contrast is required. Therefore, SFO can significantly increase the interaction rate when compared with traditional screens. The light conversion efficiency (quantum efficiency) was measured to be about 6-7 % which is low compared with the Gadolinium Oxysulfide phosphor screens' quantum efficiency of 18 %. However, the quantum efficiency is a material dependent factor and can be increased by selecting different scintillating materials. The major disadvantage of scintillating fiber optics is its low transmission efficiency that is around 3 %, and 6 % if a reflector is used. The low transmission efficiency results in the low detector efficiency of SFO.

Usually, the light produced and transmitted by the detector must create sufficient charge in the photodiode to be detected. This depends on the detector efficiency and the responsivity of the photodiodes to the scintillating light. The scintillating spectrum of the SFO matches the

response of photodiodes and the coupling between SFO and photo-devices is not a problem. However, the light generated by the SFO may not be sufficient for the direct detection by photodiodes due to the low detector efficiency of the SFO. Because the problem is caused by the low light production and not radiation interaction rate, low detector efficiency was compensated for by light amplification, without effecting other parameters.

When spatial resolution is not a critical factor, the choice between SFO and phosphor screens is made on the basis of which requires the least amount of time to satisfy the interaction rate and light output. SFO will always have a greater interaction rate and can satisfy the requirement of number of interactions in less time than phosphor screens. However, depending on how many more interactions are required in a SFO, to also satisfy the light output determines its performance in comparison to phosphor screens that always have greater light output per interaction.

A previous imaging study [45] shows that using a 420 kV x-ray source, a commercial Gadolinium Oxysulfide phosphor screen based system performs better than a Terbium Oxide scintillating fiber optics based system without image intensifier stages. However, when a 3 MeV x-ray source was used, the interaction rate advantage of the SFO detector overcame its inferior light output efficiency when compared to the phosphor screen.

An experimental study to determine the inherent spatial resolution of SFO was completed. The study indicates that the scintillating fiber

optics exhibits spatial resolution of 8-9 lp/mm for low energy photon imaging and the spatial resolution does not decrease for thicker detectors.

An experimental study to determine the spatial resolution for high energy photon imaging was also carried out. Unfortunately, there were unknown errors in the high energy experiment and in the data for estimating SFO performance for high energy photon imaging. It was found however that the spatial resolution of the SFO degraded from 8-9 lp/mm to 4-5 lp/mm when the x-ray energy changed from 200 keV to 6 MeV and 18 MeV. However, light smear due to the backscattering of x-rays was observed. This backscattering is a dominant factor in degrading the measured spatial resolution. Because the backscattering can be eliminated or reduced by good shielding, the design of shielding will be very important for high energy, high spatial resolution photon imaging. It should also be noted that even when the backscattering is included, spatial resolution of 10 mm SFO equals the spatial resolution of a 0.4 mm regular phosphor screen for 100 keV x-rays.

The practical feasibility of the SFO as image detectors is evaluated using the Scantech DADR system. The detector system is a combination of a scintillating fiber optical detector and a light amplifier, so the light output is not a critical requirement. The performance of SFO compared with three types of commercial phosphor screens was based on five system characteristics. The system spatial resolution and SNR are superior and large area contrast is comparable to those observed in phosphor screens. However, the system sensitivity is lower than that of phosphor screens, which is obviously because of the low detector

efficiency of SFO. The SFO images of electronic assemblies met the requirements of ASTM 801E#A.

An analytical detector performance model was implemented using the ITS Monte Carlo code, software packages and computer programs. The Monte Carlo code calculates the energy deposition in different fibers in the SFO bundle considering secondary scattering. The computer software calculated the transform of the line spread function to the modulated transfer function. The computer programs evaluate the transmission properties of the SFO. Some of the modeling parameters are determined by the experiments described in Chapters 3 and 4. Results of the modeling studies were compared with the experimental results.

The simulations also compared the performance of SFO with phosphor screens. Because the model simulates the characteristics of the scintillating fiber optics of different components and of different thickness for different photon detection, it is useful for detector design and performance prediction.

The simulation results shows that the SFO's spatial resolution is dominated by the performance of the EMA rather than the diameter of the fiber or secondary scattering. Currently, the absorption coefficient of EMA of the SFO is about 5 %, and an improvement of another 5 % will offer significant improvement in terms of spatial resolution. It is recommended the quality of the EMA be improved. Larger fibers may make it easy to produce better EMA. The choice between the fiber diameter and the absorption coefficient of the EMA should be optimized.

The simulation also shows that the spatial resolution will not be significantly decreased as the photon energy increases and as the detector thickness increases, which supports the conclusion of the physics analysis in Chapter II. If this is the case, the spatial resolution degradation in the high energy experiment is mainly caused by the stray light generated by the backscattering and the SFO detector will exhibit excellent spatial resolution for high energy photon imaging.

The imaging capability of the SFO is quantitatively investigated using the figure of merits (FOM). The FOM is a relatively simple method which combines the spatial resolution and signal to noise ratio. The simulation indicates that for high energy photon detection the imaging capability can be enhanced by increasing the thickness of scintillating fiber optical detectors.

7.3 Recommendation for Further Research

The research in scintillating fiber optics and their applications as image detectors has been completed. It is hope that the methods and results in this research will aid in further investigation in improving SFO's properties and in evaluating their applications. The following are suggestions for further studies.

(1) The capabilities of a SFO detector can be optimized depending on its application. For example, the fiber diameter should be optimized according to the requirement for spatial resolution and the absorption ability of EMA.

(2) Other forms of SFO can be investigated to solve the low light transmission efficiency defect. For relative low energy applications in which a thin detector is employed, crystal core and air cladding could be used. In this combination the transmission efficiency will be greatly improved. For low spatial resolution application in which the fiber diameter is relatively fat, a reflector around fibers may be used instead of the EMA.

(3) Research in the application of scintillating fiber optics in high energy photon imaging should be completed. Their operation and the design of detector shielding should be characterized.

(4) The radiation damage on scintillating fiber optics should be also investigated.

LIST OF REFERENCES

- [1] D. A. Fry, et al., Design Considerations for an Industrial Diode Array Digital Radiography System, ANS Transactions, Vol. 56, Supplement 3 (1988), pp 70
- [2] E. Bigler and F. Polack, Scintillating Optical Fiber Array for High-resolution X-ray Imaging over 5 keV, Appl. Opt., Vol. 24, No. 7, April 1, 1985, pp. 994-997
- [3] R.M. Polichar, High Resolution Prototype Real-Time Radiography System, AFWAL-TR-86-4061, Ohio, Nov. 1986
- [4] R. Ruchti, et al., A Scintillating Glass, Fiber-optic Plate Imaging System for Active Target and Tracking Applications in High Energy Physics Experiments, IEEE Transactions. Nuclear Science, NS-30, No.1, Feb., 1983, pp. 40-43
- [5] R. Ruchti, et al., Scintillating Glass, Fiber-optic Plate Detectors for Active Target and Tracking Applications in High Energy Physics Experiments, IEEE Transactions. Nuclear Science, NS-31, No.1, Feb., 1984, pp. 69-73
- [6] D.S. Gamiles, Modeling High Energy X-Ray and Gamma Ray Response of Scintillating Fiberoptics, MS Thesis, Ohio State Univ., 1987, pp 45-49
- [7] L. Reiffel and N. S. Kapany, Some considerations on Luminescent Fiber Chambers and Intensifier Screens, Rev. Sci. Instr., Vol. 31, No. 10, Oct. 1960, pp 1136-1142
- [8] V. I. Nikanorov, et al., Otchet Laboratorii Yadernykh Problem OIYaI, Report of Laboratory of Nuclear Problems, Joint Institute of Nuclear Research, May, 1957

- [9] L. T. Kerth, Radiation Laboratory, Univ. of California, Internal Memo, Nov., 1957
- [10] M. Conversi and A. Gozzini, Nuovo cimento 2, 1955, pp 189
- [11] Barsanti et al., Proc. CENR Symposium High Energy Accelerators and Pion Phys, 2, 1956, pp 56
- [12] S. R. Borenstein and R. C. Strand, Scintillating Fiber Optics, IEEE Transactions. Nuclear Science, NS-29, No.1, Feb., 1982, pp. 402-404
- [13] S. R. Boremstein and R. C. Strand, Process in the Development of Scintillating Optics Fibers, IEEE Transactions. Nuclear Science, NS-31, No.1, Feb., 1984, pp. 396-398
- [14] Michihiro Ito, et al., CsI(Na) Scintillation Plate with High Spatial Resolution, IEEE Transactions. Nuclear Science, NS-34, No.1, Feb., 1987, pp. 401-405
- [15] M. Bourdinaud and M. Laguesse, Monte Carlo Simulation and First Experimental Results of a New X-ray Detector Using Fluorescent Optical Fibers, 7th Symposium on Radiation Measurements and Applications, Ann Arbor, Michigan, 21-24, May 1990
- [16] Workshop on Scintillating Fiber Detector Development for the SSC, Nov. 14-16, 1988, Fermi Laboratory, Batavia, IL, Vol. 1,2
- [17] R. Ruchti, et al., Scitillating Glass, Fiber-optic Plate Detectors for Active Target and Tracking Applications in High Energy Physics Experiments, IEEE Transactions. Nuclear Science, NS-32, No.1, Feb., 1985, pp. 590-594
- [18] R. Ruchti, et al, A Cerium Glass Fiber-optic Active Target for High Energy Physics Experiments, IEEE Transactions. Nuclear Science, NS-33, No.1, Feb., 1986, pp. 151-154

- [19] B. Abbot, et al, Scintillating Fibers and Waveguides for Tracking Applications, IEEE Trans. Nuc. Sci. Vol. 38, No.2, April 1991, pp 441-445
- [19] K. Eggert et al., Proc. 6th Topical Workshop on Proton Antiproton Collider Physics, Achen, 1986
- [20] J. Alitti, The Design and the Construction of the Scintillating Fiber Detector for the UA2 experiment, IEEE Trans. Nuc. Sci. Vol. 38, No.2, April 1989, pp 29-34
- [21] G. J. Feldman et al, 4π DETECTOR, Proceeding of the 1984 Summer Study on the Design and Utilization of SSC, Editors R. Donaldson and J. G. Morfin, June 1984
- [22] M. D. Petroff, High Energy Tracking using Scintillating Fibers and Solid State photomultipliers, IEEE Transactions. Nuclear Science, NS-34, Feb., 1987, pp 525
- [23] K. Kuroda et al., Proposal to Upgrade the SCIFI Detector, LAPD, Annecy-Ce-Vieux, France, 1988
- [24] D. Puseljic, et al, Liquid Capillary Scintillating Detectors, IEEE Transactions. Nuclear Science, Vol. 37, No.1, April 1990, pp. 139-143
- [25] M. Atac, et al, High Resolution Scintillating Fiber Gamma Ray Detectors for Medical Imaging, 1990 IEEE Nuclear Science Symposium, Oct. 22-27, Arlington, VA, pp 1128-1130
- [26] R.C. Chaney, et al, Simulation of Scintillating Fiber Gamma Ray Detectors for Medical Imaging, IEEE, 1990 IEEE Nuclear Science Symposium, Oct. 22-27, Arlington, VA, pp 1099-1101
- [27] Varian ER 210TM Real Time Radiography System, Equipment Specifications, Varian Associates, Inc., Radiation Division
- [28] J. A. Halbleib, et al. ITS: the Integrated TIGER Series of Coupled Electron/Photon Monte Carlo Transport Codes, SAND 84-0573, Nov., 1984

- [29] S.R. Elliot, The Physics of Amorphous Materials; Longman Group Limited; 1983
- [30] D.L. Grison, Nature of defects and defect generation in optical glass, SPIE; 541, 1985, pp 38-59
- [31] E.J. Friebele, et al, Optical Fiber Waveguides in Radiation Environments, II. Nuclear Instruments and Methods in Physics Research; B1, 1984, pp 355-369
- [32] A. Yariv, Optical Electronics, 3th Edition, CBS College Publishing, New York, 1985
- [33] F.A. Hopf and G.I. Stegeman, Applied Classical Electrodynamics, John & Sons, Inc. 1885
- [34] Helmut F. Wolf, Handbook of Fiber Optics: Theory and Applications, Garland Publishing, Inc., 1979
- [35] E.G. Neumann, Single-Mode Fibers Fundamentals, Vol 57, Theodor Tamir Ed., Springer-Verlag, 1988
- [36] Kittel, Introduction to Solid State Physics, fifth edition, John Wiley & Sons, Inc., 1976, pp 187
- [37] R. Elliot, The Physics of Amorphous Materials, Longman Group Limited, 1983, pp 178
- [38] R.N. Brecewell, The Fourier Transform and Its Applications, McGraw-Hill Book Company, New York, 1965
- [39] William H. Beyer, Standard Mathematical Tables, 26th Edition, CRC press, Inc. Boca Baton, Florida, 1983
- [40] Albert Macovski, Medical Imaging Systems, Prentice-Hall, Inc., 1983
- [41] R. Singh, Digital Signal Analysis for Mechanical Systems, ME870 Class Notes, OSU, Aut. 1988

- [42] D.W. Miller, Nuclear Instrumentation, NE741 Class Notes, OSU, 1987
- [43] H.A. Bethe, Moliere's Theory of Multiple Scattering, Phys. Rev. 89,1255 (1953)
- [44] S. Goudsmit and J.L. Saunderson, Multiple Scattering of Electrons, Phys. Rev., 58, 36-42 (1940)
- [45] D. Fry, Evaluation of the Detector Subsystem and Modeling of Diode Array Digital Radiography, MS Thesis, Ohio State Univ., 1987
- [46] H.A.Kramers, On the theory of x-ray absorption and of the continuous x-ray spectrum, Philos. Mag., 836(1923) pp 46
- [47] John M. Boone, The three parameter equivalent spectra as an index of beam quality, Med. Phys. 15(3), May/Jun 1988, pp304-310
- [48] William Lowry Thompson, Gamma-ray and Electron Transport by Monte Carlo, Ph.D Dissertation, Univ. of Virginia, 1984, Chap.2, pp 19-20
- [49] A. B. Chilton, Principles of Radiation Shielding, Prentice-Hall, Inc., New Jersey, 1984
- [50] R.D. Evans, The Atomic Nucleus, McGraw-Hill, New York (1955), pp 100-102
- [51] J.B. Birks, The Theory and Practice of Scintillation Counting, The Macmillan Company, New York, 1964
- [52] P. Goldberg, Luminescence of Inorganic Solids, Academic Press, New York and London, 1966
- [53] G.F. Knoll, Radiation Detection and Measurement, John Wiley & Sons, USA, 1979, Chap. 8, pp. 239-242

- [54] B.I. Stepanov and V.P. Gribkovskii, Theory of Luminescence, London Iliffe Books, LTD, 1968
- [55] W.F. Nelson, et al, Fluorescence of Tb^{3+} and $Tb^{3+}-Eu^{n+}$ in Fused Silica, Proc. 3rd Rare Earth Research, Gordon and Breach, New York
- [56] C. W. Bates and R. Morwood, Information Transfer Using Thin Transparent Membrans, App. Opt., Vol. 12, Sept. 1973, pp 2185
- [57] B.G. Streetman, Sold State Electronic Devices, 2nd Edition, Prentice-Hill, Inc., Englewood Cliffs, New Jersey, 1980
- [58] E. A. J. Marcatili, Dielectric Rectangular Waveguide and Directional Coupler for Integrated Optics, The Bell System Technical Journal, Sep., 1969, pp. 2071-2102
- [59] D. Miller, The Optical Coupling between Scintillating Fiber Optic Imaging Guides for Radiographic Applications, MS Thesis, Ohio State Univ., 1987. PP 17-37
- [60] M. Kuroda and S. A. Collins, Coupling Light Radiated by Atoms into Dielectric Waveguides, Unpublished paper, Dept. of Electrical Engineering, OSU, 1989
- [61] L. B. Felsen, RAYS, Models and Beams in Optical Fiber, Opt. and Quan. Electronics, Sep., 1977, pp 189-195
- [62] R. J. Potter, Transmission Properties of Optical Fibers, J. Opt. Soc. America, Vol 51, No. 10, Oct. 1961, pp 1079-1089
- [63] Synergistic Detector Designs, 2438 Wyandoth, Bldg A, Mtnview, CA, 94043
- [64] Schott Fiber Optics, 122 Charlton St., Southbridge, MA, 01550
- [65] B.K. Utts, Recent Advances in the Extrusion of Alkali Halide Crystals for Scintillation Radiation Detector, Presented at the 2nd London Conference on Position Sensitive Detectors, Sep 4-7, 1990, Imperial College, London

- [66] R. J. Potter, Transmission Properties of Optical Fibers, Opt. Soc. America, Vol 51, No. 10, Oct. 1961, pp 1079-1089
- [67] A. J. Davis, et al, Attenuation and Effective efficiency in small fiber waveguides, Workshop on Scintillating Fiber Detector Development for the SSC, Fermi Laboratory, 1988, pp 117
- [68] Personal Communications, The Ontario Cancer Institute, 500 Sherbourne Street, Toronto/Canada, M4x 1k9, June 6, 1984
- [69] personal communication with Mr. Walter Siegmund of Schott Fiber Optics
- [70] Schneiders, N. J, and S. C. Bushong, "Single-Step Calculation of the MTF from the ERF," Medical Physics, 5(1), Jan/Feb, 1978, pp 31-33
- [71] Sashin, D., et al., "Radiography Apparatus," United States Patent No. 4,179,100, 1979
- [72] D. A. Fry, et al., "Design Considerations for an Industrial Diode Array Digital Radiography System," ANS Transactions, Vol. 56, Supplement 3 (1988), pp 70-72
- [73] ASTM standard designated E 801-85, "Standard Practice for Controlling Quality of Radiographic Testing of Electronic Devices"
- [74] Hua Shao, Don W. Miller and C. Robert Pearsall, "Scintillating Fiber Optics for X-ray Radiation Imaging," 7th Symposium on Radiation Measurements and Applications, Ann Arbor, Michigan, April 21-24,1990
- [75] Levy Hill Laboratories Ltd., Cheshunt, Herts, England
- [76] E.J. Friebele, et at, Overview of radiation effects in fiber Optics, SPIE; 1985, 541, 70-88

- [77] A. Bross, Radiation Damage to Scintillators, Workshop on Scintillating Fiber Detector Development for the SSC, Nov. 14-16, 1988, Fermi Laboratory, Batavia, IL, pp 741-768
- [78] A. H. Cherin, AN INTRODUCTION TO OPTICAL FIBERS, McGraw - Hill Book Company, America, 1983
- [79] M. Born, et al, Principles of Optics, Sixth edition, Pergamon Press, England, 1980, pp 38-40
- [80]. H. Shao, et al, Scintillating Fiber Optics and Their Application in Radiographic Systems, IEEE Trans. Nuc. Sci, Vol 38, April 1991, 845-857
- [81] K. Rossmann, "The spatial Frequency Spectrum: A Means for Studying the Quality of Radiographic Imaging System", Radiology, Vol. 90, No. 1, Jan., 1968, pp 1-13
- [82] K. Rossmann, "Comparison of Several Methods for Evaluating Image Quality of Radiographic Screen-Film Systems," Am. J. Roentgenol, Vol 97, July 1966, pp 772-775
- [83] F.R. Whitehead, "Quantitative Analysis of Minimum Detectable Lesion to Background Uptake Ratios for Nuclear Medicine Imaging System", Vol 1, Medical Radionuclide Imaging, IAEA proceeding, pp 409-432
- [84] J. W. Steidley, Imaging Capabilities of Germanium Gamma Cameras, PhD Dissertation, OSU, 1977
- [85] D. M. Green and J. A. Swets, Signal Detection Theory and Psychophysics, John Wiley and Sons, Inc. New York, (1966)
- [86] A.G. Schultz, "quantitative Assessment of Scanning System Parameters", J. Nucl. Med. 11,2, pp61-68, (1970)
- [87] K. Rossmann and R.D. Moseley, "Measurement of the Input to Radiographic Imaging Systems", Radiology, Vol. 92, No. 2, Feb. 1969, pp 265-271

- [88] A. Notea, "Evaluating radiographic systems using the resolving power function", NDT International, Vol. 16, No. 5, Oct. 1983, pp 263-270
- [89] A. Ginsburg, Amer. J. Optometry & Physiological Optics, Vol. 59, No. 1, Jan. 1982
- [90] Collimated Holes, Inc., 460 Division Street, Campbell, CA 95008
- [91] S. A. Collins, Class Notes, EE917 Advanced Topics in Electrical Engineering, 1989
- [92] Menu of Siemens Mevatron KD-2, Siemens Medical Laboratories, Inc. Concord, California, 1987
- [93] K. Rossmann, et al, Measurement of the Line Spread-Function of Radiographic Systems Containing Fluorescent Screens, Journal of the Optical Society of America, Vol. 54, No. 2, Feb. 1964, pp 187-190
- [94] D. Miller, H. Shao et al., "Fiber Optics in Radiation Environments", EPRI Project Report, RP-2614-69, 1991
- [95] MATLAB package, User's Guide, The Math Works Inc., January 1, 1991