To my family – Mom, Dad, and Mike
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Over the last twenty five years, medical imaging utilizing ionizing radiation has significantly increased and as a result an individual is more susceptible to receive a higher exposure over the course of his or her lifetime. Pregnant female diagnostic studies have followed this trend as there has been a 25% increase in Computed Tomography (CT) imaging studies from 1997 to 2006.\textsuperscript{1} An increase in the number studies coupled with newer scanner technology, which has shown to maintain if not raise radiation exposure for a single exam, have resulted in significant increase in individual and collective effective dose.\textsuperscript{2, 3} Most pregnant women that undergo CT exams will incur a fetal absorbed dose of less than 50 mGy over the course of their pregnancy. This level of absorbed dose is the recognized threshold for concern, as it is less than what is needed to induce abnormalities such as growth and mental retardation, which are common effects documented in the epidemiological studies of the Japanese bomb survivors. However, this level of exposure is comparable to radiation workers who by law require individual monitoring. For an occupational worker who is declared pregnant, the limit is 5mSv per gestation or 0.5mSv in any one month of
pregnancy. Given the sensitivity of young children to radiation compared to adults, this discrepancy in monitoring needs to be addressed in some form.

Using anatomically accurate reference physical models and recently developed advanced dosimetry techniques, the absorbed dose to the fetus and embryo will be evaluated at multiple gestation stages from clinical CT studies performed at the conventional imaging protocols using a modern scanner. The results of these studies are compared to both current biological data to validate or disprove claims about radiation safety in diagnostic radiology, and current clinical fetal/embryonic dosimetry methods. Most of these resolved uncertainties stem from the effectiveness of tube current modulation imaging and the accuracy of dose calculations using scanner reported DLP values. The dissertation is also intended to advocate any additional imaging precautions, or necessitate new and advanced techniques of monitoring conceptus doses from clinical CT studies.
CHAPTER 1
SUMMARY OF ESTIMATING CONCEPTUS DOSE, EVALUATING STOCHASTIC RISK, AND THE EFFECTS OF RADIATION ON UNBORN CHILDREN

Clinical Dosimetry and Risk

Unshielded irradiation delivered to an unborn child in diagnostic radiology is avoided completely or limited to situations where the mother’s health is severely jeopardized, as evidenced by the practice in clinical medicine of ALARA (As Low As Reasonably Achievable). The concept of ALARA is especially applicable to Computed Tomography (CT) studies, which cumulatively account for a significant portion of the population’s annual effective dose received in diagnostic studies. In order to minimize the fetal dose, clinical protocols are adjusted and additional precautions are utilized. Some of the modified imaging techniques include: using a filter that adds hardening to the beam, collimating the x-ray fan beam to the diameter of the patient, collimating the z-axis beam width to prevent wasted radiation, reducing the time-current product directly, minimizing the scan length to only including critical anatomy, and the use of tube-current modulation scanning techniques. The extent to which these safeguards reduce the fetal dose in the later stages of pregnancy is based on the estimated conceptus dose studies in the literature.

Before examining the risk of fetal doses from imaging studies, it is worthwhile to look at the recommended dose limitations that are currently established for pregnant females who may be potentially exposed to radiation. Currently, the National Council on Radiation Protection and Measurements (NCRP) recommends that the effective dose to the embryo or fetus of an occupational worker not exceed 0.5 mSv per month of gestation. The International Commission on Radiation Protection (ICRP) claims that no more than 2 mSv of equivalent dose should be delivered to the surface of the
pregnant woman’s abdomen during her pregnancy. In the case of diagnostic imaging, where low kVp x-rays are being used, these quantities can also be represented as 0.5 and 2 mGy of absorbed dose, respectively. It should be noted that in x-ray dosimetry the equivalent dose (in mSv) and absorbed dose (in mGy) are numerically equivalent as the radiation weighting factor, \( w_R \), for x-rays and electrons, is equal to unity. Although these regulations are enforced for pregnant occupational workers, they do not apply to medical imaging. The benefit of a diagnosis or treatment that utilizes radiation will outweigh the risk from receiving radiation, especially if that risk is stochastic in nature.

**Clinical CT Dosimetry**

The helical nature of the x-ray source in multi-slice computed tomography requires a unique metric to evaluate x-ray exposures from a specific clinical CT study. Application-specific dose metrics have been introduced as a measurement that can be easily performed that accesses x-ray tube output and generally correlates with dose for the specific imaging modality. Since the 1970’s, Computed Tomography Dose Index (CTDI) has been the standard metric for quantifying the x-ray tube output from a scanner operating under a specific imaging technique. The CTDI is defined as the average measured air kerma to the center of a cylindrical acrylic phantom. For a multislice CT scanner capable of simultaneously scanning (N) of nominal slice thickness (T), the measured air kerma from a single full rotation along a line parallel to the axis of rotation, over a length of 100 mm, can be theoretically expressed as:

\[
\text{CTDI}_{100} = \frac{1}{NT} \int_{-50}^{+50} D(z) \, dz 
\]  

(1-1)

The instrumentation to measure the integral of the single rotation dose profile, \( D(z) \), is a 100-mm ion pencil chamber which measures an integrated exposure reading
over its active length (100mm). The integral in Equation 1-1 simplifies to the measured exposure by the chamber corrected for air kerma using the f-factor multiplied by the chamber’s active length. There are two types of CTDI phantoms, a head (16-cm diameter), and body (32-cm diameter) phantom. Both phantoms are acrylic cylinders that have measurement points for the ion chamber on the periphery of the phantom at orthogonal angles and a single location at the center. For helical scanning, the definition of CTDI was modified to account for beam overlap or beam spacing, by accounting for the exam pitch. Measurements are taken at both the center and periphery, and weighted accordingly, and adjusted based on the pitch, $p$, to find the CTDI volume (CTD$\text{D}_{\text{vol}}$). CTD$\text{D}_{\text{vol}}$ represents the average dose across the area of the central scan plane (z=0). It is defined in terms of $CTDI_{100}$ in Equation 1-2.

$$CTDI_{\text{vol}} = \frac{1}{3}(CTDI_{100, \ c} + 2 \ CTDI_{100, \ p}) \quad (1-2)$$

The calculation of $CTDI_{\text{vol}}$ takes into account multiple locations within the phantom plane vs. a single axial measurement. However it is really only a truly accurate measure if the pencil chamber can capture the primary beam and the scatter tails. Therefore with regards to wide beams the accuracy of this method becomes uncertain. To provide a more realistic quantity of the total radiation exposure, the Dose Length Product (DLP) was introduced to account for the scanned length. Thus for a multi-slice helical scanner, with a scan length ($l$), the DLP is defined as:

$$DLP = CTDI_{\text{vol}} \ast l \quad (1-3)$$

The scanner reports the DLP value based on the beam profile and output tables generated by the manufacturer, and the specified technique for that exam: kV, mA,
pitch, rotation time, and scan length. The values for DLP and CTDIvol are estimated values and not intended to explicitly represent any kind of patient dose. Therefore, it is common practice to measure CTDIvol and DLP for different scan techniques. These measurements are not used to calibrate the scanner, but merely to provide a secondary measurement of radiation output. These values are very crude estimates, and are already outdated since there is no way to accurately measure CTDIvol for a 16 cm wide beam (320 slice scanner).  

The CTDI methodology in quantifying radiation exposure is unique to CT. Other dose quantities include entrance skin air kerma (ESK) in radiography and the kerma air product (KAP) in fluoroscopy. Both of these quantities are measured free in air using a designated ion chamber. All of them are independent of each other and require much further corrections to obtain any type of patient specific absorbed dose. One way to compare patient doses for various imaging modalities is to utilize the effective dose (ED). Therefore, the DLP-to-effective dose coefficient (ED/DLP) was developed to evaluate CT patient doses with respect to other imaging modalities and to provide organ dose data to evaluate stochastic risk. In a journal publication by Huda, DLP values for different scanners and effective doses calculated by Monte Carlo dosimetry codes were implemented into a software program to estimate effective dose. Table 1-1 shows a comparison of three available dosimetry software programs that uses these parameters to estimate the effective dose from CT scans performed frequently in the clinic. All three software packages used the mathematically constructed stylized phantoms. These computational phantoms are different from any real patient since they are based on standard height, weight, organ size and location. Also, the mean effective dose
and DLP are given for a reference sized individual weighing approximately 70 kg. The study acknowledges that when scanning patients whose size differs from 70 kg, different correction factors will have to be used if reliable ED values are to be obtained. The ED/DLP conversion factors overall provide a relative risk quantity but do not give an absolute risk assessment, lending itself as a useful method for determining the risk from exam-to-exam.

There are many limitations when using the CTDI methodology for patient dosimetry especially when applied to newer generation scanners. Most obvious, is the geometry of a typical patient is not very similar to a 32-cm diameter cylinder. The dose profile within a slice is highly dependent on the diameter of tissue within that slice. This was explained in detail by a series of studies correlating radiation absorbed dose in CT to effective diameter. As previously stated, CTDI cannot be quantified for volumetric scanners which utilize beams collimated to widths (FWHM) greater than 100 mm. The third limitation is that CTDI underestimates dose. Most clinical scans lengths are greater than 100mm, and the scatter contribution from the regions outside the volume occupied by the chamber and phantom will contribute to the accumulated dose. At scan lengths greater than 25 cm, the dose approaches a limiting equilibrium. CTDI_{100} underestimates the equilibrium dose by a factor of 0.6 on the central axis and 0.8 on the periphery. CTDI_{vol} places a higher emphasis on the periphery doses which may help offset the underestimated equilibrium dose since most organs lie in the middle of the patient. When it relates to exams that utilize tube current modulation the quantity CTDI_{vol} becomes more of an issue. The nature of CTDI_{vol} is the average dose to a phantom at the central scan axis (z=0). All of the slices are assumed to be similar in
geometry and the scan parameters remain constant throughout the entire scan, an accurate estimate of dose can be estimated. The issue when quantifying CTDI in the traditional sense is that the spatial dose profile of the scan changes from the modulated tube current along the scanning axis. Boone and Dixon showed that three completely different TCM protocols can yield similar CTDIvol values.\textsuperscript{12} The dose profile cannot be captured by the ion chamber from a single rotation scan at a constant mA, therefore the value of CTDI cannot be accurately quantified from just a single measurement. Even if it could evaluated, the parameter would be spatially dependent and therefore a series of values would be required corresponding to a slice, and then matched with the target organ in order to correlate with absorbed radiation dose.

**Clinical Fluoroscopy Dosimetry**

As a means of offering a comparison between CT and most of the other imaging modalities, a section on fluoroscopy was included. The fluoroscopy tube head contains a Roentgen-area product (RAP) meter mounted below the collimator to measure the intensity of the x-ray beam striking the patient. The RAP meter is sensitive to exposure rate, time the machine is imaging the patient (fluoroscopy time), and area of the beam. Other imaging parameters such as kVp, magnification mode of the image intensifier, and source-to-patient distance are typically not recorded by the RAP meter. Skin dose is the one of the main concerns in fluoroscopy due to long exposure times, leading to significant reported skin doses. The RAP meter is a useful tool for estimating the skin dose as the dose-area product can be mapped onto the surface of the patient. Applying correction factors, the skin dose may be calculated.

The x-ray source in fluoroscopy is delivers radiation in projections relative to the patient. The conceptus dose is calculated for a single projection using a manual
technique facilitated by using The Book Handbook of Selected Tissue Doses for Projections Common in Diagnostic Radiology. The output of the x-ray tube for a given tube potential and filtration, or HVL, is measured by an ion chamber 100 cm from the x-ray source. The parameter is measured in ionization (mR) per time-current product (mAs). The entrance skin exposure (ESE) for each projection can then be calculated using Equation 1-4.

\[ ESE = mAs \times \left[ \frac{mR}{mAs} \right] \times \frac{100^2 \, cm^2}{SSD^2} \] (1-4)

Where the source to surface distance (SSD) and mAs are measured or given for each projection. The handbook lists tissue dose per ESE at different projections for different organ sites tabulated with various beam HVL’s. While this method accounts for many parameters, patient characteristics such as size, shape, and body composition are ignored along with the collimated field size. These imaging parameters significantly contribute to the back scattered radiation which adds additional exposure.

**General Risk Assessment**

The effective dose (ED) is a dose quantity related to evaluating stochastic effects. It is the standard metric when estimating cancer risk from a non-uniform radiation exposure. The effective dose is calculated by totaling the tissue-weighted organ doses of specific organs in the body that are most susceptible to stochastic effects from ionizing radiation. The formula, originally defined in ICRP 60, for calculating effective dose from x rays is shown in Equation 1-5.

\[ D = \sum_{tissue,T} w_T D_T \] (1-5)
Where \( w_T \) is the tissue-weighting factor, and \( D_T \) is the equivalent dose (expressed in mSv) derived from the average organ absorbed dose multiplied by a radiation-weighting factor specific to the incident radiation energy and type. Photons and electrons are always assigned a radiation weighting factor of 1 therefore \( D_T \) can be expressed as the absorbed dose averaged over the entire organ.

ICRP 103\(^{14}\) is the revised publication that incorporates updated the tissue-weighting factors for all risk-contributing organs. These factors are a relative risk assessment of each organ with respect to the others. The breast tissue weighting factor for example is 0.12, but the brain is assigned a value of 0.01. By definition it would take approximately twelve times the radiation to produce the same stochastic event in the brain, as it would for the breast. The tissue weighting factors are derived from a uniform whole body exposure. Weighting is based on individual organ’s relative contribution to the total detriment.

According to the ICRP, the effective dose is defined independent of age and gender. The definition of the effective dose lacks in specificity since the relative radiosensitivity of organs differs greatly with age. The case studies of the Chernobyl accident revealed a very high incidence in thyroid cancer for children versus a moderate prevalence in adults. Other examples include the higher risk of leukemia for children and high incidence of lung cancer relative to breast cancer for women over 60 versus that of an infant. The high contribution of the breast to the total detriment significantly impacts the effective dose calculation. The breast weighting factor is significantly overestimated for men and underestimated for women. Additionally, there is a significant difference in the amount of breast tissue between a 5 year old and a 30 year
old woman. Another issue or common criticism of the effective dose is that it incorporates physical and hereditary effects, non-fatal and fatal cancers, into one factor for each organ over the entire population. Each of these end points should correspond to a different weighting factor, and any attempt to apply a weighted average of these endpoints is both arbitrary and subjective. Given the apparent differences in both anatomy and development between a fetus and an aged adult and the partial irradiation characteristic of these diagnostic studies, the effective dose is not an effective metric for evaluating stochastic risk for a developing fetus from CT exposures. As a result most radiobiology studies report fetal radiation absorbed dose rather than effective dose.

**BIER VII**

The most comprehensive risk-management study is by the BEIR VII committee which published its most recent report in 2005. The report published relative risk coefficients for a variety of organ specific solid tumor sites, whole body, and leukemia, relating low-levels of ionizing radiation to cancer and genetic effects. These empirical values are age and gender dependent, and account for elapsed time after the exposure event. The detriment is evaluated for three separate endpoints: incidence, fatality from radiation-induced cancer, and severe hereditary effects. The term “excessive relative risk” corresponds to the additional risk contributed from a radiation exposure producing a radiation weighted absorbed dose, D. The term absolute relative risk combines the relative risk to the baseline, or unexposed lifetime risk. The functional form of the dose-response curve is described by Equation 1-6, where $ERR$ denotes the excessive relative risk.

$$ERR = B_s D exp[\gamma e^*] \left(\frac{a}{60}\right)^\eta$$  \hspace{1cm} (1-6)
The model is derived from the linear-no-threshold (LNT) model. The term $\beta_s$ is the sex dependent coefficient, $D$ is the radiation-weighted dose (solid tumor) or effective dose (leukemia) in Sv. The variable, $e^*$ is a parameter dependent of the age of exposure, while $a$ is the attained age, and lastly $\gamma$ and $\eta$ are empirically tabulated factors.

The BEIR VII report found that health effects from very low doses would be difficult to evaluate due to contributions from other spontaneous factors, which are also statistical in nature, interfering with extrapolation of the data to these low-dose levels. This “noise threshold” described in the report is stated to be 100 mSv, although the risk is expected to follow a linear trend with no threshold. Therefore any measured dose under 100 mSv would impose statistical limitations on identifying the root cause, but the risk-dose relationship will presumably remain linear.

**Studies Evaluating Conceptus Dose**

A study by Jaffe\textsuperscript{17} was conducted to correlate the relationship of absorbed fetal dose in the early first trimester with CTDIvol when performing tube current modulation scans with variable noise index settings on a physical anthropomorphic reference female phantom. Measurements were obtained using an anthropomorphic female phantom, and segmenting the uterine wall to function as the dose point. Dose was measured using a metal oxide semi-conductor field effect transit (MOSFET) detector, along with the reported scanner values for CTDIvol. Low dose protocols at 140 kVp were parameterized for a 16 slice multi-detector CT (MDCT) GE scanner (Lightspeed 16, GE Healthcare, Madison, WI). The results showed a linear correlation of CTDIvol with absorbed fetal dose. The reported absorbed dose ranged from 9.25 to 37.7 mGy. Studies by McCollough\textsuperscript{18} obtained similar measurements, also claiming conceptus dose
from single acquisition CT does not in normal circumstances exceed a nominal value of
35 mGy as seen in Table 2. Although Jaffe did evaluate the effects of TCM on a first
trimester pregnant female, the modulation of the beam under these circumstances is
minimal given the relatively uniform effective diameter of the abdomen and pelvic
regions across the phantom. Moving from a first trimester pregnant female to a second
trimester should induce a significant modulation in the scanner output, which could
more accurately demonstrate the use of CTDIvol in TCM studies.

Another study by Hurtwitz et al.\textsuperscript{19} evaluated the radiation dose to the fetus at 0
and 3 months of pregnancy from a MDCT scanner. The radiation source was a 16-slice
MDCT scanner operating at standard protocols. Measurements were taken using a
commercially available anthropomorphic female phantom modified to represent a newly
pregnant patient and a separate phantom that was 3 months pregnant. The physical
phantom was used with a MOSFET dosimeter. The reported fetal doses were as
follows: renal stone protocol, 8-12 and 4-7 mGy; appendix protocol, 15.2-16.88 and 20-
40 mGy; and pulmonary embolus protocol, 2.4-4.7 and 0.6-0.66 mGy.

A study by Damilakis et al.\textsuperscript{20} used the Monte Carlo N-Particle (MCNP) radiation
transport code to evaluate the CT conceptus dose from two clinical scanners, Siemens
Sensation 16 and Sensation 64 MDCT. Four mathematical phantoms simulated women
at 0, 3, 6, and 9 months of gestation. Measurements were separately verified using an
anthropomorphic phantom with a set of Lucite rings and TLD dosimeters. The data was
used to formulate a general formulism for calculating the conceptus dose from the free
air CTDI value, abdomen perimeter, and conceptus distance. The most
comprehensive study, by Angel et al.\textsuperscript{21}, generated 24 patient-specific voxelized models
from a cohort of 31 pregnant patients, spanning a range of gestational ages, whom previously underwent clinical CT abdomen and pelvic exams. This study accounted for the diverse position and location of the fetus and discovered a very important correlation. Fetal dose in CT correlated to both the maternal perimeter and centroid of the fetus’s depth within the mother. Dosimetry was however limited to studies performing single acquisition abdomen/pelvis exams.

**Radiobiology**

In addition to stochastic effects, developmental effects are of great concern when assessing damage to an irradiated fetus or embryo. Developmental effects have a proven dose threshold, unlike carcinogenesis which is probabilistic in nature. Animal studies, epidemiological studies from the Japanese survivors, and cohort studies pertaining to medical exposures have provided information on both stochastic and developmental effects from ionizing radiation. From this information developmental effects have been categorized into three categories: lethal effects, malformations, and growth disturbances without malformations. Lethal effects are induced by radiation before or immediately after implantation of the embryo into the uterine wall, and increasingly higher doses thereafter. Embryo or fetal death can be categorized as either prenatal (before birth) or neonatal (at birth). Malformations are characteristic of interference or disturbances of organ development. Growth disturbances can occur at any time over the course of the pregnancy, but malformations are characteristic of radiation damage during early stages of gestation.
Gestational Periods

In order to better categorize risk from radiation, Russell and Russell divided the total developmental period into three stages: preimplantation, organogenesis, and fetal period.\textsuperscript{22}

Preimplantation

Preimplantation with regard to associative risk from radiation exposure is best characterized as "all or nothing", inferring a high risk of prenatal death, and a very low risk of any abnormalities. This period extends from the time of fertilization to the implantation of the embryo into the wall of the uterus. This period of time usually ranges from 8 to 10 days, but is ultimately defined by the zygote transforming into a blastocyst. The embryo is highly sensitive to radiation because most cells are simultaneously entering mitosis and there is a small existing cell population. Most radiobiological data from this period comes from animal studies. Since irradiation during this period is all or nothing and death is prenatal it often goes undetectable or considered a failure to implant and therefore difficult to make to determine a root cause. Animal studies collectively have placed an overall LD50 at 1\text{Gy}.\textsuperscript{23} This value ranges considerably, (1.5 to 0.3 \text{Gy}) due to gene expression and effects of radiosensitivity caused by the cell cycle. The human zygote is engaged in mitotic activity when it divides from a single cell into 2 cells. Mitotic activity, which is considered to be the most sensitive stage of the cell cycle, occurs 2 to 6 hours post conception. Post fertilization, the zygote is least sensitive to radiation because gene expression has yet to manifest itself. It is commonly accepted that malformations cannot be induced by radiation during this time period. Animal studies show that malformations exist, but do not accurately demonstrate a dose threshold.\textsuperscript{24}
**Organogenesis**

Organogenesis, which is the second phase of gestation, is the period during which most organs are actively developing. It is during this period that congenital abnormalities associated with structural malformations are most evident. This time frame begins when the embryo has been implanted in the uterine wall and lasts until the 6th week post conception. Although a majority of deaths are neonatal, prenatal death is prevalent in early organogenesis. In early organogenesis, cells are at their most differentiating stage, as the embryo is rapidly increasing in size. Embryos exposed to radiation during early organogenesis show the greatest intrauterine growth retardation. However, growth retardation is recoverable, both in size and weight, before or after birth.25

**Fetogenesis**

The last phase of gestation is the fetal period. During this period structures are already formed and are actively growing in size. Common effects include damage to the hematopoietic system, liver, and kidneys. The gonads are also vulnerable to irradiation during this time period. The early fetus exhibits the largest degree of permanent growth retardation.26 The fetal period is characterized by of the transformation of an embryo into a fetus. Dekban27 conducted a survey that looked at instances of pelvic x-ray irradiation of pregnant women and concluded that irradiation after 30 weeks of gestation is not likely to produce gross structural abnormalities leading to a permanent handicap in early life, but functional disabilities were possible. Irradiation of the fetus between 16 and 25 weeks of gestation may lead to a mild degree of microcephaly, mental retardation, and stunting of growth. Irradiation between 4 to 11 weeks leads to severe abnormalities of many organs in most children, most frequently
stunted growth, microcephaly, and mental retardation. Abnormalities associating with
the eye, skeleton, and gonads are also frequent, but in literature are not as frequent as
the listed neurological disorders.

**Specific Developmental and Carcinogenetic Effects**

Most malformations induced by low levels of radiation are cephalic, while other
types are deemed limited. This is the result of migration of brain cells within the Central
Nervous System (CNS) during the early development of the CNS. Data from the
Japanese bomb survivors reveal microcephaly, mental retardation, and growth
retardation as the primary effects.\(^{28}\) Radiation related mental retardation is limited to
exposure during 8-25 weeks post conception, while most pronounced between 8-15
weeks. This time frame represents a period in which there is greatest proliferation of
immature neurons and their migration from the ventricular proliferative zones to the
cerebral cortex. The Japanese survivor data also demonstrates a dose threshold,
which studies have determined to fall between 100 to 300 mGy.\(^{29}\) Note that these
values for absorbed embryonic/fetal dose are estimated. Another study concluded an
IQ decline of 21 points/Gy for the 8 to 15 week period, and 13 points/Gy for the 16-25
week period.\(^{30}\) ICRP\(^{7}\) concluded that radiation-induced mental decrement is
deterministic with a threshold related to the minimum shift in IQ that can be measured.
Using the previous data and fitting it to a linear –threshold model, the threshold would
result in about 50 mGy, which hasn’t been validated. Existing data has provided no
clear evidence for a dose threshold with respect to IQ loss.

Microcephaly, decreased head diameter, is observable only for exposures
occurring in the time frame 1 to 16 weeks post fertilization, with no significant excess in
risk seen after 16 weeks. This morphological finding was most prevalent in the
Japanese bomb survivor’s data. Microcephaly is not prevalent during the 16-25 week period when mental retardation can occur. As is the case, mental retardation doesn’t appear visible in the 1-8 week time period. The highest risk of mental retardation occurs when the cerebral cortex is being formed, which occurs after 8 weeks. Cells killed before this period can cause small head size without affecting neuron development because the neurons that lead to the formation of the cerebrum are not yet sensitive to radiation, unlike the glial cells which provide structure for the brain which tend to be susceptible for depletion. Microcephaly is a specific growth defect, and is related to the number of damaged or killed cells. No current evidence supports a precise dose threshold.\textsuperscript{31} BEIR VII claims that the minimum dose at which this effect has been observed in humans falls between 100 to 190 mGy.

Carcinogenetic risk from in-utero irradiation has been well documented and generally accepted as a liner-no-threshold response. Human epidemiological studies have shown that in-utero exposure to diagnostic x-rays is associated with increased risk of leukemia and, to a lesser extent, solid tumors.\textsuperscript{32} The root cause of solid tumors is difficult to ascertain due to the extended latent period of the cancer. These tumors are usually visible during the period of highest incidence which coincides in the later stages of human life, further adding to the difficulty in distinguishing the mutagenic agent between ionizing radiation and a genetic predisposition. Gestational age has been shown to play a large role when accessing stochastic risk from prenatal irradiation. Animal data has shown a relative increase in risk during the early stages of gestation.\textsuperscript{33} This could be the result of scientific evidence that associates replication errors caused by radiation in proliferating cells. The embryo/fetus undergoes multiple cell divisions.
during its lifetime, making it more sensitive to neoplasms. Similarly, the incidence of
cancer is dose-rate and LET dependent, which has also been well documented.\textsuperscript{31} One
study, by the Children’s Cancer Group, found no evidence for an overall increase in risk
for diagnostic in-utero radiation.\textsuperscript{34} The authors credit this finding to a decrease in
diagnostic x-ray imaging involving pregnant females over the past decade. The OSCC
(Oxford Survey of Childhood Cancers) study\textsuperscript{35}, which is the most comprehensive
diagnostic radiation study, calculated a cancer risk estimate of 640 excess cancers per
10,000 people per Gy and a 95% confidence interval between 410-1000 excess
cancers. However, the Japanese bomb survivors have shown a lower excess cancer
incidence probability. The Japanese bomb survivor’s data was unable to incorporate
leukemia into its findings, due to the early onset of leukemia. It is also contested by
Wakeford and Little\textsuperscript{36} who suggest that the number from the OSCC study should be
three times lower, due to artifacts in its most recent years. Cohort studies have shown
that there is almost no risk to prenatal diagnostic radiation exposure. Although these
studies are very precise experimentally, they do not represent a large population of
childhood cancer cases and hence have very broad confidence intervals.

**ICRP 90 Conclusions**

The ICRP 90\textsuperscript{25} publication is a document that reviewed experimental animal data
on the in-utero effects of medical radiation exposures, and evaluated a series of human
studies concerning the in-utero risk of cancer incidence and effects on the developing
brain. Regarding carcinogenesis and malformations, the following conclusions were
drawn for a prenatal irradiation below 100 mGy of low LET radiation:

- In retrospect of the induction of malformations, there are gestational age-
dependent patterns of in-utero radiosensitivity with maximum sensitivity being
  expressed during the period of major organogenesis, and on the basis of animal
data, there exists a true dose threshold of around 100 mGy for the induction of malformations.

- With regard to neurological effects, the A-bomb survivor data strongly supports a dose threshold of 300 mGy for severe mental retardation during the most sensitive prenatal period (8-15 weeks post conception). The associated data on IQ loss is difficult to predict, and a non-threshold dose response cannot be excluded, however IQ loss following the exposure of under 100 mGy would be of no practical significance.

- Regarding cancer risk, it is acceptable to correlate life-time cancer risk following in-utero exposures to irradiation in early childhood.

The National Cancer Institute developed a risk assessment calculator based on the risk models proposed by 2006 BEIR VII publication with additional risk models developed by NCI for eight cancer sites, as well as additional cohort studies to update the thyroid and breast models. The calculator employs a set of techniques to evaluate the estimated lifetime risk from a distribution of potential lifetime risks taking into account statistical uncertainties associated with the risk model parameters and subjective uncertainties due to a number of assumptions. Table 1-3 lists a series of organs and tumor sites along with the calculated mean relative risk of a 0 to 1 year old male or female exposure to a uniform 50 mGy radiation absorbed dose. From Table 1-3 the most prevalent types of cancer are leukemia, and cancer from the liver and thyroid. The last two organs are weighted individually according to ICRP 103 as 0.04 of the total detriment. In addition to those organs and the stomach, the other sites are equal at producing a carcinogenetic event. With this mind, it further demonstrates the lack of specificity in using the effective dose with regard to the fetus.

The term relative risk is somewhat misleading given that the baseline risk of childhood cancers is extremely low and the lifetime risk of cancer over the course of one’s lifetime is very high. Therefore, estimating cancer risk is very cumbersome due to
the difficulty in ascertaining the cause of cancer from radiation or genetics due to similar attained ages of carcinogenetic expression which is occurs mostly in the later part of life. In addition the epidemiological and cohort studies that use these models show wide confidence margins and apply a small statistical sampling to a large population as shown by the vast confidence intervals in determining ERR from the last column in Table 1-3. Although not listed The Health Physics Society\textsuperscript{39} states that with regard to risk assessment:

Radiological risk assessment, particularly for radiogenic cancer, currently is only able to demonstrate a consistently elevated risk in the intermediate- and high-dose groups of the studied populations. Cancer and other health effects have not been observed consistently at low doses ($< 100$ mGy), much less at the even lower doses ($< 10$ mGy) typical of most occupational and environmental exposures. Consequently, in order to estimate radiation risk in the low-dose region, observed health effects in the higher-dose regions are extrapolated to the low-dose region by using a variety of mathematical models, including the linear, no-threshold model (with a correction for dose and dose rate).

With regard to risk assessment the radiobiology is completely different between a developing fetus and adult, and therefore what is applicable to one may not necessary to be applicable to other. A fetus and embryo are physiologically undergoing more cell differentiation, migration, and proliferation than an adult, making it more susceptible to low-levels of ionizing radiation and therefore posing a higher risk.\textsuperscript{40}

**Summary of Effects**

Developmental effects induced by radiation exposure to a developing embryo in CT are possible. The two most prevailing and common effects documented in literature are mental and growth retardation. The minimum dose threshold for mental retardation is stated to fluctuate between 100 and 200 mGy. This taking place between the 8-16\textsuperscript{th} week gestational period. Any exposure under 100 mGy falls below the established
threshold for the formation of any congenital abnormalities. With regards to deleterious effects, the American College of Obstetricians and Gynecologists\textsuperscript{41} published the following opinion: “Women should be counseled that x-ray exposure from a single diagnostic procedure does not result in harmful fetal effects. Specifically, exposure to less than 5 rad [50 mGy] has not been associated with an increase in fetal abnormalities or pregnancy loss.” A report conducted by the Health Protection Agency of the United Kingdom found that the natural childhood cancer in the UK is reported as 1 in 500, and a fetal exposure as small as 10 mGy was claimed to double the baseline risk.\textsuperscript{42} From Table 84 in ICRP 84, 50 mGy would result in no increased risk of abnormalities and an ERR of 3.0\%.\textsuperscript{43} Sufficient biological evidence is not available to support a claim that a single CT examination, or a conceptus dose under 50 mGy, would jeopardize the development of a fetus or embryo. Finally, the effective dose is a completely inadequate method in evaluating low dose stochastic effects, as it compounds the error in accessing risk to an unborn child from a diagnostic radiation exposure. It also overestimates and underestimates the detriment contribution from individual organs at low doses. From Table 1-3, the breast, colon, and lung contribute a lot less to the overall risk than the thyroid and liver.

The 50 mGy benchmark has been well documented to exceed the fetal dose received from most CT examinations, on the account of the vast majority of routine single-phase acquisition computed tomography studies deliver less than 35 mGy to the uterus. To achieve a conceptus dose greater than 50 mGy, a pregnant patient would have to undergo multiple CT exams. Practically, no medical clinician would allow this to happen if he or she had knowledge the patient was pregnant, leading to the scenario
where the patient and doctor are unaware of pregnancy, and the patient is subject to several examinations. The rate at which pregnant women are unintentionally exposed to radiation is largely unknown. One study involving trauma patients concluded that 2.9% of the women between the ages of 15 and 40 years admitted to the trauma unit were found to be pregnant. Of the pregnant women who were admitted, 11% were incidental pregnancies, of which 8% were newly diagnosed.44

**Research Objective**

For the purposes of patient dosimetry risk it can be assumed that any exposure resulting in less than 5 mGy, the occupational limit for the declared pregnant worker, can be assumed to be acceptable since this limit is determined without accounting for the clinical benefit of receiving a diagnostic test to the overall welfare of the mother. This manuscript is not intended to justify alternative imaging methods or promote a dose limitation. The research in this study simply attempts to advocate the need to monitor radiation exposures from clinical CT for females that are or may be potentially pregnant, in the same way we monitor radiation workers. In addition, this research will examine the accuracy of current methods of estimating doses with the intent on validating their place within the clinic.

The information presented in this chapter highlights the significance of monitoring and evaluating embryonic/fetal doses in CT. The overall objective of this dissertation is to advance the physical tools and methods used currently to quantify conceptus doses associated with low levels of ionizing radiation used in current CT procedures than the studies presented in this work. The results will be used to provide a framework that will allow medical personnel to conduct clinical CT studies so that they may result in minimal risk to the women and unborn child. Each specific aim is highlighted below:
1. Construct a fiber-optic dosimeter that is independent to variation in depth and energy, and can easily be integrated with anthropomorphic phantoms. In addition, the dosimeter is able to measure absorbed dose to a variety of bodily tissues. Characterize the dosimeter to verify that it maintains a dose rate and angular independence, and high sensitivity and reproducibility to radiation seen with water-equivalent FOC dosimeters.

2. Characterize the overall error to using anthropomorphic phantom materials for dosimetry studies, and computationally evaluate the sensitivity of measuring dose with these materials in comparison to their tissue counterparts. This goal includes the development of a new methodology to expedite the phantom construction process, eliminate the intermittent anatomical structures that results from building the computational phantom from axial slices, and create dose points within the phantom in the sagittal and coronal anatomical planes allowing for dose points in all three anatomical planes.

3. Develop a dosimetry methodology for measuring absorbed dose to anthropomorphic phantoms that incorporates the linear fiber-optic dosimeter and the energy independent dosimeter. The linear fiber-optic dosimeter is a device that has been characterized to exhibit minimal variation in sensitivity to ionizing radiation along its axis, making the ideal dosimeter for measuring dose in one dimension. By combining multiple s the dose across the entire volume may be evaluated more accurately.

4. Fetal/embryo dose measurements, using each pregnant female phantom and the dosimeters discussed above performed using a clinical CT scanner. The scan protocols administered were selected based on their clinical significance and the level of radiation delivered to the abdomen and pelvis.

5. Use collected data to benchmark modern methods of estimating the conceptus dose received from clinical CT studies examined in aim 4.

The following chapters of this dissertation are dedicated to addressing each of the aims discussed above in greater detail. Each chapter corresponds to a specific numbered project above and gives the pertinent background information, experimental methodology, and results. An extra chapter was added to summarize the effects of these results and how they pertain to the clinical use of CT on pregnant and potentially pregnant females. Another chapter was incorporated to address the issues of using fiber-optic dosimeters in high energy applications specifically external beam photon and electron radiotherapy.
Table 1-1. Comparison of effective dose calculated from CT dose software for female from multidetector CT (MDCT). Protocols selected based on European guidelines.10

<table>
<thead>
<tr>
<th>Dosimetry Software</th>
<th>Effective Dose (mSv)</th>
<th>Head</th>
<th>Chest</th>
<th>Abdominal</th>
<th>Pelvic</th>
</tr>
</thead>
<tbody>
<tr>
<td>WinDose</td>
<td>1.8</td>
<td>8.8</td>
<td>12.1</td>
<td>10.5</td>
<td></td>
</tr>
<tr>
<td>ImpactDose</td>
<td>2.2</td>
<td>11.0</td>
<td>13.0</td>
<td>14.0</td>
<td></td>
</tr>
<tr>
<td>CT - Expo</td>
<td>2.6</td>
<td>13.8</td>
<td>16.4</td>
<td>16.5</td>
<td></td>
</tr>
<tr>
<td>Coefficient of Variation (%)</td>
<td>17.5</td>
<td>22.1</td>
<td>16.4</td>
<td>22.2</td>
<td></td>
</tr>
</tbody>
</table>

Table 1-2. Conceptus doses from a sample of common clinical CT studies. Bottom three protocols indicated nominal doses exceeding 2 mGy.18

<table>
<thead>
<tr>
<th>Examination Protocol</th>
<th>Fetal Dose [mGy]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head CT</td>
<td>0</td>
</tr>
<tr>
<td>Chest CT</td>
<td>0.2</td>
</tr>
<tr>
<td>CTA (chest)</td>
<td>0.1</td>
</tr>
<tr>
<td>Routine Abdominal</td>
<td>4</td>
</tr>
<tr>
<td>Routine Abdomen/Pelvis</td>
<td>25</td>
</tr>
<tr>
<td>CTA (chest thru pelvis)</td>
<td>34</td>
</tr>
</tbody>
</table>
Table 1-3. Lifetime risk from a 50 mGy acute exposure to a 0-1 year old child.

<table>
<thead>
<tr>
<th>Organ/Site</th>
<th>Gender</th>
<th>Excess # of cancers</th>
<th>Baseline # of cancers</th>
<th>Total # of cancers</th>
<th>ERR (confidence interval)</th>
</tr>
</thead>
<tbody>
<tr>
<td>All Organs</td>
<td>M</td>
<td>1990</td>
<td>48290</td>
<td>50280</td>
<td>3.96% (2.00% - 6.30%)</td>
</tr>
<tr>
<td>All Organs</td>
<td>F</td>
<td>3350</td>
<td>41230</td>
<td>44580</td>
<td>7.51% (4.33% - 12.32%)</td>
</tr>
<tr>
<td>Brain/CNS</td>
<td>M</td>
<td>69.7</td>
<td>661</td>
<td>730</td>
<td>9.55% (2.25% - 23.68%)</td>
</tr>
<tr>
<td>Brain/CNS</td>
<td>F</td>
<td>19.6</td>
<td>537</td>
<td>557</td>
<td>3.52% (0.93% - 8.29%)</td>
</tr>
<tr>
<td>Breast</td>
<td>F</td>
<td>819</td>
<td>13176</td>
<td>13994</td>
<td>5.85% (3.17% - 8.51%)</td>
</tr>
<tr>
<td>Colon</td>
<td>M</td>
<td>242</td>
<td>4079</td>
<td>4321</td>
<td>5.60% (2.91% - 9.25%)</td>
</tr>
<tr>
<td>Colon</td>
<td>F</td>
<td>165</td>
<td>4200</td>
<td>4364</td>
<td>3.78% (1.62% - 7.21%)</td>
</tr>
<tr>
<td>Leukemia</td>
<td>M</td>
<td>177</td>
<td>939</td>
<td>1116</td>
<td>15.86% (1.65% - 57.30%)</td>
</tr>
<tr>
<td>Leukemia</td>
<td>F</td>
<td>155</td>
<td>722</td>
<td>876</td>
<td>17.69% (1.22% - 69.01%)</td>
</tr>
<tr>
<td>Liver</td>
<td>M</td>
<td>77.6</td>
<td>833</td>
<td>910</td>
<td>8.53% (1.35% - 29.90%)</td>
</tr>
<tr>
<td>Liver</td>
<td>F</td>
<td>43.3</td>
<td>349</td>
<td>392</td>
<td>11.05% (0.83% - 44.54%)</td>
</tr>
<tr>
<td>Lung</td>
<td>M</td>
<td>234</td>
<td>7931</td>
<td>8165</td>
<td>2.87% (1.25% - 5.28%)</td>
</tr>
<tr>
<td>Lung</td>
<td>F</td>
<td>552</td>
<td>6326</td>
<td>6878</td>
<td>8.03% (3.81% - 15.64%)</td>
</tr>
<tr>
<td>Mouth</td>
<td>M</td>
<td>38</td>
<td>1427</td>
<td>1465</td>
<td>2.59% (0.29% - 6.04%)</td>
</tr>
<tr>
<td>Mouth</td>
<td>F</td>
<td>36.5</td>
<td>683</td>
<td>719</td>
<td>5.08% (1.17% - 11.19%)</td>
</tr>
<tr>
<td>Ovaries</td>
<td>F</td>
<td>74.5</td>
<td>1445</td>
<td>1519</td>
<td>4.90% (1.23% - 12.17%)</td>
</tr>
<tr>
<td>Prostate</td>
<td>M</td>
<td>139</td>
<td>16385</td>
<td>16523</td>
<td>0.84% (0.00% - 5.32%)</td>
</tr>
<tr>
<td>Stomach</td>
<td>M</td>
<td>121</td>
<td>1095</td>
<td>1217</td>
<td>9.94% (0.76% - 39.59%)</td>
</tr>
<tr>
<td>Stomach</td>
<td>F</td>
<td>148</td>
<td>683</td>
<td>831</td>
<td>17.81% (1.59% - 73.35%)</td>
</tr>
<tr>
<td>Thyroid</td>
<td>M</td>
<td>134</td>
<td>376</td>
<td>510</td>
<td>26.27% (5.01% - 77.72%)</td>
</tr>
<tr>
<td>Thyroid</td>
<td>F</td>
<td>742</td>
<td>1070</td>
<td>1812</td>
<td>40.95% (7.08% - 137.88%)</td>
</tr>
<tr>
<td>Uterus</td>
<td>F</td>
<td>46.3</td>
<td>3241</td>
<td>3287</td>
<td>1.41% (0.00% - 4.71%)</td>
</tr>
</tbody>
</table>

* all cancers listed are mean number of cancers for a population of 100,000 people within a 90% uncertainty range
** acute 50mGy exposure to a person who is born in 2014 and exposed in 2014
***confidence intervals calculated from given upper and lower bounds for excess # of cancers and total #
CHAPTER 2
HIGH ENERGY COUNT RATE EFFECTS AND BACKGROUND REMOVAL ISSUES IN PLASTIC SCINTILLATING FIBER-OPTIC DOSIMETRY

Counting Issues in Fiber-Optic Dosimeters Used in Radiotherapy Dosimetry

The stem effect is traditionally a term coined for the additional charge created outside the sensitive region of an ionization chamber or diode and subsequently measured contributing an excessive signal. In most situations, the ionization chamber’s stem is usually in the radiation field, therefore a correction is not needed when the calibration is performed with the stem completely in the field. For fiber-optic detectors this value is significantly dependent on the amount of optical cable in the radiation field, which is highly variable. With respect to fiber-optics used in radiation detection, the stem effect refers to the natural fluorescence of optical fibers under exposure to ionizing radiation leading to additional signal contribution. A portion of the light generated within the optical fiber is collected adding to the signal from the scintillator. Not properly accounting for the optical fiber’s light contribution can impact the accuracy of measurements especially under varying x-ray field sizes.

Plastic scintillating dosimeters (PSDs) have many characteristics that prove to be very beneficial for radiotherapy dosimetry. Their sensitive volume, polystyrene, has been determined through Monte Carlo methods to be energy independent for x-ray photons above 500 keV. In addition, they can be fabricated into miniature dosimeters making them very useful for internal in vivo dosimetry. Plastic scintillators are very sensitive to ionizing radiation and can therefore be made with little material to produce an adequate signal. Their main drawback is the optical fiber component, which is an integral component of the PSD system, produces a prompt and substantial signal when
exposed to ionizing radiation. This is signal is significant enough to severely affect the accuracy of radiation dose measurements, especially in the megavoltage energy range.

Hyer examined the applicability of using PSDs for radiotherapy dosimetry. His characterization of these detectors, and the work of Beddar, have demonstrated excellent dose linearity and reproducibility, as well as an adequate response to varying pulse repetition rates. However, the findings uncovered a dependence related to the radiation field length, when using the background subtraction method. This dependence hasn't been discussed in great detail or observed in other publications. The literature has somewhat successfully demonstrated the use of PSD’s under small field sizes for IMRT and Stereotactic radiosurgery modalities as well as the achievements of chromatic filtering using capable devices such as CCD cameras. This chapter will examine how the background signal from an optical fiber affects the quality of dosimetry for a detection system using a plastic scintillator, photomultiplier tube array, and no use of any chromatic filtering or spectral analysis techniques, which recently have proven to be quite successful.

The Stem Effect

Cerenkov Radiation

The stem effect is much more observable at megavoltage energies than at kilovoltage energies because of Cerenkov radiation. Cerenkov radiation is electromagnetic radiation emitted when an electron passes through a dielectric medium at a velocity greater than the speed travels in the medium. As a result of this phenomenon, more background counts are detected by the dosimetry system. The threshold for Cerenkov radiation has been estimated to above 178 keV in PMMA and 144 keV in polystyrene. In radiotherapy applications, using background subtraction
as a Cerenkov removal technique has shown some success.\textsuperscript{53, 54, 59} This method is ultimately limited by the spatial resolution of the two fibers under the presence of a strong dose gradient perpendicular to the fibers, as its ultimate success is dependent on equal irradiation of each optical fiber. The collection of light produced from Cherenkov radiation is highly dependent on the direction of the charged particle. In contrast to light produced from scintillators, in which light is emitted isotropically, Cherenkov photons are emitted in the direction of the particle’s path. The light is released in a cone defined by an angle ($\theta$) which is defined as:

$$\theta = \cos^{-1}\left(\frac{1}{\beta n}\right)$$  \hspace{1cm} (2-1)

The variable $\beta$ is the ratio in velocity of the particle to the speed of light, and $n$ is the index of refraction of the medium. All light produced inside the optical fiber cannot be collected by the PMT because the mechanism for channeling light through a narrow dielectric (via total internal reflection) is limited to the angle defined by the numerical aperture of the optical fiber. Light striking the walls of the fiber will be totally internally reflected, provided that the incident angle at each reflection is greater than the critical angle:

$$\sin \theta_c = \frac{n_{\text{clad}}}{n_{\text{core}}}$$  \hspace{1cm} (2-2)

Figure 2-1 illustrates the geometry of the optical fiber in relation to the incident ionizing particle and the totally internally reflected light propagating through the core of the optical fiber. The light collection cone of the optical fiber represented by $\varphi$ can be expressed in radians as:
\[
\varphi = \left( \frac{\pi}{2} - \sin^{-1}\left( \frac{n_{\text{clad}}}{n_{\text{core}}} \right) \right)
\]  

(2-3)

In most cases the optical fiber is positioned orthogonally to the direction of the incident radiation. High energy charged particles typically interact with matter in soft coulombic interactions that do not affect the trajectory of the charged particle. Therefore, in the case of superficial megavoltage electrons and, more so for protons, the Cerenkov radiation generated in the optical fiber will lie outside the acceptance cone of the optical fiber. However in the case of x-rays, which interact mostly through Compton scattering, interactions will produce charged particles in the axially direction that fall under the acceptance cone of the optical fiber. Figure 2-1 presents the trajectory of a charged particle inside an optical fiber along with the Cerenkov cone with respect to the acceptance cone of the optical fiber. For the Eska optical fiber (Eska CK-20, Mitsubishi Rayon Co, Tokyo, Japan), the core and cladding index of refraction is 1.49 and 1.403 respectively, which gives a cone with a vortex angle \((\varphi)\) 0.303 radians, translating to a solid angle of 0.954 radians.

**Color Centers**

Irradiation of plastic scintillating fibers and optical fibers has shown to affect their optical properties.\(^{60-62}\) The effect is a temporal and/or permanent loss in the transmission of light. The reduction in light transmission is attributed to the formation of color centers inducing an increase in optical absorption properties of wavelengths in the visible/UV light spectrum. A study examined the color formation by irradiation for a variety of polymers and found that all exhibited a shift in the UV cutoff and the formation of an absorption tail which can extend into the visible light range.\(^{63}\) The formation of color centers are characterized into two types: permanent and annealable, of which the
latter is a recoverable process achieved by the release of trapped radicals via migration of free oxygen or radical-radical reactions without the presence of air.\textsuperscript{64, 65} The annealing process through oxygen for polystyrene can be characterized on the order of hours up days post irradiation, and is sensitive to variation in dose rate.\textsuperscript{66, 67} Any residual damage after this time period can be considered permanent. For polystyrene it was observed that transmission damage observed during irradiation is considerably larger than permanent damage especially for wavelengths in the visible light spectrum.

The transient effects from recombination originate from the formation of electron-hole pairs or free radicals generated by ionizing radiation within the lattice structure of the fiber.\textsuperscript{68} Most of these free radicals recombine; the remaining free radical species are trapped inside imperfections within the lattice and may be released through other mechanisms, most notably oxygenation as discussed above. It is these trapped electrons and free radicals that affect the time-dependent optical transmission properties of waveguides. Following irradiation, electron-hole pairs are formed by ionizing radiation within the fiber initiating a transient absorption effect that results in immediate loss of light transmission. At lower doses of radiation the transmission of a fiber will increase towards the pre-irradiation value. The time between the transient absorption and the intrinsic absorption is known as recovery. The recovery kinetics and transient absorption was evaluated by Mattern et al.\textsuperscript{61} for four types of known radio-resistant fibers: polystyrene, PMMA, Ge-doped silica, and pure vitreous silica. The recovery kinetics for each fiber type follows a power-law behavior with around 90% recovery taking place within 1ms. The data was obtained using small radiation pulses at very high dose rates well exceeding those experienced clinically. However, the data
strongly supported claims that the recovery kinetics of all the fibers examined after a transient absorption appear to be independent of dose, irradiation type, and wavelength which was evaluated between 500 to 800nm.  It was also noteworthy that the peak transient absorption for PMMA and silica occurred at 400 nm and 550 nm respectively, and about 4 times lower in magnitude for the silica fiber thus making the PMMA core fiber more sensitive to optical absorption of Cerenkov photons.

**Ideal Stem Response to Radiation for a Pair of Optical Fibers**

The majority of the light reaching the PMT are photons released by the scintillator, while the remaining contributing portion is from the natural fluorescence of the optical fiber from either natural luminescence or Cerenkov mechanisms. Luminescence is a result of de-excitation of orbital electrons or recombination of free radicals with electron-hole pairs within the fiber. Alternatively the energy deposited may be in the form of heat. This additional amount of light may be accounted for by adding an optical fiber of equal length and placed parallel to the signal fiber. This positioning would allow the fiber to experience the same x-ray intensity and energy spectra, and simplify the relationship between the stem counts in each fiber to a single factor. The number of detectable steam counts at any position along the optical fiber can be expressed fundamentally by Equation 2-4.

$$N(x) = I(x) * S * \varepsilon_{fiber} * \left[ \frac{\mu_{en}}{\rho} \right]_{fiber} * A_{sect} * \rho * dx$$  \hspace{1cm} (2-4)

where $N$ is equal to the number of counts detected from irradiation at position $x$. $S$ is the scintillation efficiency of the glass fiber. $I(x)$ is the x-ray beam intensity along the fiber’s axis. $A$ is the cross section area of the fiber and $L$ is the length of the fiber being
irradiated. The term $\varepsilon_{fiber}$ is the fraction of light produced in the fiber that is detected by the PMT, and can be further represented as a product of other variables.

$$\varepsilon_{fiber} = \varepsilon_{accept} \varepsilon_{transmit} \varepsilon_{PMT} \quad (2-5)$$

Where:

- $\varepsilon_{accept}$ represents the fraction of light photons produced in the scintillator that successfully travel through the optical fiber. This parameter is often associated with the acceptance angle characterized by the numerical aperture of the optical fiber.

- $\varepsilon_{transmit}$ represents the transmission efficiency of light reaching the end of the optical fiber. Signal loss is attributed to light attenuation within the optical fiber.

- $\varepsilon_{PMT}$ represents the detection efficiency of light photons detected by the PMT. This value is variable among other PMT devices and is dependent on the wavelength of the light photon.

These additional terms represent the relative amount of light produced in the optical fiber that reaches the photon counter. For two optical fibers placed contiguous and parallel to each other under uniform x-ray field, the assumption of equal amounts of radiation is absorbed by each fiber at the same location is valid. Under a uniform radiation field, the relationship between the quantities of detected stem counts in each fiber can be expressed using Equation 2-6.

$$\frac{N_1}{N_2} = \frac{\varepsilon_2}{\varepsilon_1} \quad (2-6)$$

Using Equation 2-6, the reference fiber can be calibrated to measure the stem counts in the scintillator fiber. The ratio of light collection efficiency may be obtained by placing the optical fiber inside a radiation field with the scintillator shielded and placed outside the primary beam while irradiating the optical fibers at varying energies and field sizes to determine the relationship in their responses. Equation 2-6 also describes an
ideal response for the background subtraction method, which the measured counts in both fibers should correlate linearly with length of fiber irradiated.

**Count Rate Effects**

**Radiation Delivery Mechanisms of Clinical Radiation Generators**

**Continuous sources**

Some clinical x-ray sources are delivered from a beam current incident to a high Z medium target that produces bremsstrahlung radiation. These are typically radiological kilovoltage x-ray tubes and orthovoltage tubes. Long-lived radionuclides also generate a constant level radiation over a short time span. The dose rate is controlled by the current flowing into the target and the distance the absorber is from the high-Z target. Under fixed conditions a detector placed within the x-ray beam experiences a steady-state source of constant intensity and produces a relatively stable count rate.

**Linear accelerator (LINAC)**

Bremsstrahlung radiation emitting from a LINAC is generated in pulses from the pulsing of the magnetron supplying the accelerator waveguide with oscillating microwaves for duration of several microseconds at a frequency of 3,000 MHz. A typical Varian Trilogy CLINAC calibrated to deliver 1 cGy per Monitor Unit (MU) at an SSD of 100 cm and a depth of 1.5 cm for a 6 MV beam produces an estimated instantaneous dose rate of 77.22 Gy/s (0.00028 Gy per pulse). This radiation source can be represented by the time-dependent sketch in Figure 2-2. The larger pulses are separated by the pulse repetition frequency (1/f) which is expressed in terms of monitor units per minute, which clinically can fall between 100 and 1000 MU/min. Standard calibration protocols are set to deliver 100cGy per 100 monitor units at 100cm source-
to-detector distance (SSD) to an ionization chamber immersed inside a water phantom located at the depth below the surface corresponding to the maximum dose.

**Dead Time Effects from a PMT Detection System**

Potential counting losses may arise at high count rates from dead time behavior of scintillation counting systems. Medical imaging devices such as SPECT and PET that use sensitive detectors with high absorption efficiencies require performance evaluation at high count rates for image distortion effects and loss of image contrast. The energy-to-light conversion efficiency of polystyrene in the sub 100 keV range is relatively small compared to inorganic scintillators that are more sensitive to low energy photons due to their high atomic number and inherently higher photon yields. The smaller light yield of the plastic scintillator creates less dependence on the time resolution characteristics of the photon detector, since the count rate is smaller dead time contributions are also smaller.

The count-rate linearity of a fiber-optic system is determined by the resolution time of that system, which is defined as the minimum time between successive scintillation pulses that the detection system can distinguish. This time interval may be limited by the decay rate of the scintillator, the frequency characteristics of the signal processing circuit, and the timing characteristics of the photodetector's electronics. Photomultiplier tubes (PMTs) are photon counting devices that have an exceptionally fast time response. The time response of a PMT is determined by the electron transit time. Electron transit time is the time interval between the arrival of the light photon at the cathode and its subsequent collection at the anode after being multiplied. The period between arrival and collection, however, only induces a fixed delay in the signal. The main concern with respect to dead time is the transit time spread (TTS).
TTS is the temporal distribution of electrons arriving at the anode due to differences in electron pathlength and velocity. A wide distribution may lead to potential overlapping between pulses leading to loss of signal. The concept of TTS is illustrated in Figure 1. Transit-time fluctuations in the electron transit time have two components: a chromatic dependence due to the spread of photoelectron initial velocities, and a geometric dependence caused by variation in the distance traveled by the electron from the cathode to the anode. The chromatic dependence may be offset by operating the PMT at wavelengths corresponding to its peak sensitivity. This results from a larger number of photoelectrons being produced in the cathode, offering a better sampling of photoelectrons. Commercial PMTs used for photon counting require the use of additional electronic devices that process the output pulses from the PMT. Additional integrated electronic components may further modulate the shape of the individual pulses which may lead to pulse overlapping. Therefore, it is important to evaluate the timing properties of the entire counting system rather than individual components.

**Dead Time Modeling**

Counting losses incurred due to the time resolution may potentially be modeled using one of the two statistical classifications of dead time behavior: paralyzable and nonparalyzable. It is widely recognized that scintillation camera systems have both nonparalyzable and paralyzable operating ranges. The nonparalyzable model predicts that the detector will be temporarily insensitive to radiation during the dead time, while in the paralyzable model, the detector remains sensitive to any additional events which may lead to pileup. In this scenario, two events separated by a time less than the dead time would result in neither event being detected. In the nonparalyzable case, the detector is able to detect the first event but not the second leading to saturation of
detectable counts at high count rates. The true count rate may be analytically solved for by Equation 2-7.

\[ n = \frac{m}{1 - m \tau} \]  

In Equation 2-7, \( m \) identifies the observed count rate, \( n \), the true count rate, and \( \tau \) is the dead time. However, this model explains an ideal behavior as \( \tau \) is subject to variation inevitably leading to inaccuracies, especially at high count losses given the sensitivity of \( n \) to changes in the product \( m \tau \). Since the system dead time is determined by the time resolution of the individual components of the system, any components that are not characterized as nonparalyzable may lead to a nonconventional model.

Additional considerations need to be addressed when modeling dead time behavior from pulsed sources. Knoll\textsuperscript{76} addresses three cases where the dead time (\( \tau \)) is less than \( T \), less than \( T \) but greater than \( 1/f \), and greater than \( T \) but less than the off time. In all three cases the radiation intensity is assumed constant throughout the duration of the pulse. However none of these models may apply to our condition, which is complicated by the fact that two sets of pulses on two different magnitudes of frequency are occurring due to pulsing from the magnetron and the electron gun. A model by Daams\textsuperscript{77} was derived for the apparent dead time of a detector measuring intensity of radiation which varies periodically.

\[ m_0 = n_0 (1 - n_0 \tau - \frac{n_0}{2\omega} \sin \omega \tau) \]  

The Equation 2-8 is derived for a simple periodic function where \( m_0 \) and \( n_0 \) represent the average detected and true counting rate respectively. In this case a maximum number of counts are observed which corresponds to \( \omega \tau \approx \frac{3}{2} \pi \). The
response of the detector to an oscillating source shows a linear increase in observed counts as a function of frequency unit it reaches the maxima.

**Materials and Methods**

**Detection System**

The plastic scintillating dosimetry system was constructed by Hyer and characterized for diagnostic energy dosimetry.\(^7^8\) The dosimeter consists of two cylindrical, 500 micron diameter multi-mode PMMA optical fibers (Eska CK-20, Mitsubishi Rayon Co, Tokyo, Japan), which were lengthened to 13 meters to account for LINAC vault rooms.\(^7^9\) One of the fibers was coupled to a plastic scintillating fiber element, polystyrene. An uncoupled optical fiber is utilized to account for the uncontrollable radiation induced fluorescence and Cerenkov radiation discussed above.

The scintillator was manufactured as a plastic cylindrical fiber (500 microns). Before coupling, the scintillator and coupling end of the optical fiber surfaces were polished using an Amphenol hand-puck (Ocean Optics) and 3 grits of lapping paper (Angstrom Lap, 12, 3, and 1 micrometer) to increase the number of light photons incident to the PMT (H7467, Hamamatsu Corporation, Bridgewater, NJ).\(^7^9\) The scintillator and optical fiber were manually coupled using a transparent adhesive gel (Crazy Glue) and between 1 and 2 cm of 3/64" PVC heat-shrink tubing (Fisher Scientific) to ensure a tight coupling. After coupling, the plastic scintillator was cut to 2 mm from the coupling surface, and polished. A reflector coating (Eljen Technology, EJ-510) was applied to the uncoupled end of the scintillator in order to increase the number of measurable scintillation photons.\(^8^0\) To provide structural and shield from ambient light, the two fibers were then encased in 1/16" heat shrink tubing.
Considering the transient absorption effects as a possibility for apparent quenching of the detector from high energy radiation, a gadolinium detector was constructed. Gadolinium oxysulfide (GOS) is a scintillator with a very long decay time, on the order of hundreds of microseconds, which would make the detector more sensitive by overcoming the initial absorption losses. Using the same method as above, a (GOS) scintillator was obtained in the form of a 100μm layer from a rectangular sheet of an image intensifying screen for clinical screen film mammography (Kodak MIN-R 2 Cassette, Eastman Kodak Company). The GOS was coupled to a polished optical fiber by applying a thin coat of the adhesive gel to the polished end of the optical waveguide and placing the fiber onto the slab while exerting pressure to enhance the binding process. After a short time, the optical fiber was removed along with the attached thin layer of GOS coupled to the end of the fiber.

A single PMT is required for monitoring each optical fiber. Therefore the fibers had to be separated from the heat shrink tubing to allow for connection with the PMT array. The uncoupled end of each optical fiber was mounted with a female SMA 510 connector (SMA-905) using a SMA epoxy glue(FiberFin, 1656 resin, #80 hardener). The remaining uncovered optical fiber was wrapped in heat shrink tubing. A male SMA optical fiber adapter (E5776-51, Hamamatsu Corporation, Bridgewater, NJ) was used to interface the fiber to the PMT. The counting data was buffered and transferred through a hub to a computer for readout. A USB/serial hub (UPORT 1610-8, Moxa Inc, Brea, CA) provides an efficient communication pathway between the computer and the PMT array. Each PMT and the hub are powered by 12V and 5V DC power supplies, respectively.
Sources of Radiation

The following types of radiation sources were utilized to explain and offer a comparison in the behavior of the plastic scintillation counting system at very high dose rates, as well as the ionization chamber used to accurately quantify the dose rate.

A. Varian Trilogy CLINAC (2300IX, Varian Medical Systems, Palo Alto, CA): 6 MV and 6 MeV photon and electron sources delivering radiation in pulse sequences described in Table I.

B. Clinical Proton Accelerator at the University of Florida Proton Therapy Institute (Cyclotron, ESS, IBA, Louvain-La-Nueve, Belgium). Delivers protons in a continuous beam current using a cyclotron with ESS between 25 to 300 nA.

C. Cobalt-60 irradiator. Cobalt 60 source that can be controlled using a lever system to position the source at variable distances from the detector.

D. Orthovoltage floor mounted unit (Gulmay Medical Limited, Suwanee, GA) with x-ray tube (NDI-321, Varian Medical Systems, Palo Alto, CA) selected imaging techniques between 150 and 320 kVp.

Sources labeled B, C, and D were used to characterize the dead time of the counting system using the nonparalyzable dead time model. Variable dose rates were achieved by altering the radiation current. In source B and C the radiation had enough energy to generate Cerenkov radiation, however a lesser extent in source B. The true count rate was expressed as the measured dose rate which was evaluated using the appropriate ionization chamber. The same background correction factor was used for each experiment. It should also be noted that sources B and D were equipped to control the size of the radiation field which in turn could expose the optical fibers to varying known exposure levels of radiation.

Determining the count rate from source A irradiations was difficult because measuring the count rate requires knowledge of the “beam on time” which isn’t continuous and measurable with the available equipment. Therefore the dead time
couldn’t be measured directly but, could be tested to see if the nonparalyzable model would still hold true for pulsed sources. This was done by creating a modified version of the parameter $\tau$ to include the beam on time. The dose rate was varied by changing the position of the detector or using a water phantom to take measurements and provide different penetration thicknesses. However the number of pulses was kept relatively constant since the number MU’s delivered at each position were unchanged.

**Experimental Methods**

**Stem count behavior**

Equation 2-4 explains the ideal behavior of a pair of fiber-optic cable’s response to radiation. Under the presence of a uniform radiation field, the expected value of N would be proportional to the length of cable within the field. Sources A, B, and D are all equipped with collimators to define a field size measurable at a reference location. By placing the optical fiber within and parallel to the radiation field, measurements can be taken to evaluate $I(x)dx$. Multiple readings may be acquired over a range of field sizes to determine the linearity of the background optical fiber to radiation. When considering the axial profiles of electron and proton megavoltage beams they are characterized as relatively flat, as well as diagnostic energy beams used in radiology. The spatial intensity profile of the radiation beam can be evaluated by using devices such as film and diode arrays.

**System time resolution**

The time resolution, from sources continuously irradiating the dosimeter, was evaluated by delivering a variable dose rate to the scintillator, which can be achieved by controlling the beam current. To measure the true count rate, a dedicated ion chamber was used to quantify the exposure rate and exposure time. The observed count rate
and exposure rate were calculated by averaging the net scintillation counts or exposure and dividing by the exposure time. For this method, the true count rate is expressed in terms of the exposure rate, as the two are related by a conversion factor.

The dead time was calculated by expressing the data in terms of the inverse observed count and the inverse exposure rate. Rearranging the terms in Equation 2-7, the dead time can be solved for using linear regression to find the optimal y-intercept of the inverse of the true count rate and observed count rate data:

$$\frac{1}{m} = \frac{1}{n} + \tau$$  \hspace{1cm} (2-9)

In Equation 2-9, the theoretical value of $n$ is the product of the dose rate and a conversion factor relating the number of true detected counts to the measured dose. Dose and dose rate quantities may be evaluated with an ionization chamber specific for the treatment or imaging modality. In the megavoltage energy range, this value is independent of the energy. However, fortunately for this method, only the relative value of $n$ is required to find the y-intercept. An instrument such as ion chamber is suitable to find the relative true count rate. The error associated with using this fit can be calculated using Equation 3-5 which is discussed in more detail in Chapter 3.

**Gadolinium oxysulfide and the plastic performance**

Each dosimeter was evaluated for PRF and field size dependence. The pulse rate dependence of each detector was investigated by delivering a constant dose at varying pulse repetition frequencies. Data was normalized to the lowest PRF. The field size dependence was investigated to find the effects of varying quantities of Cerenkov radiation by referencing the output factor. The response of the optical fiber is proportional to the amount of radiation irradiating the optical fiber, which can be
quantified in terms of the length of the optical fiber in the radiation field. Measurements were taken at five field sizes and normalized to the 10x10 cm field size to represent the output factor for several widely used field sizes in radiation therapy. A 5-cm slab of solid water was placed below the detector to provide backscatter. The output factor was measured using a parallel plate chamber (N23343-1754, PTW, Germany) placed in 5 cm of solid water.

**Percent depth dose**

A series of percent depth dose (PDD) curves were measured taken from various field sizes between 5x5 and 20x20 cm with a water phantom (1D Scanner, Sun Nuclear, Melbourne, FL) with a ADCL calibrated farmer chamber (Exradin A12 Ion Chamber, Standard Imaging, Middleton, WI) and associated electrometer (Victoreen’s Model 530, Elimpex, Austria). The measurements were taken in step and shoot position intervals of 1 cm from 3 to 10 cm within the phantom, not in the continuous fashion that is common to this type of measurement. Each dosimeter was shielded with a coating to prevent damage from water. For each field size, a PDD graph was created using the raw PDD and the corrected PDD data using the nonparalyzable dead-time model. The data was corrected by iterative methods to find the optimal correction factor that matched the PDD data using the least squares method. The correction factor is denoted as \( \alpha \) since the count rate could never be determined for every exposure. This is because the detection system is limited to an integration time of 10 ms. Therefore equation 2-9 is modified to evaluate \( \alpha \) for a series of equivalent pulse (magnitude and time profile) that register \( M \) total counts and \( N \) true counts.

\[
\frac{1}{M} = \frac{1}{N} + \alpha \tag{2-10}
\]
In Equation 2-10, the parameter $\alpha$ is equal to $\tau/t$ where $t$ is the equivalent time that corresponds to the average count rate from $M$ measured counts.

**Results**

**Gadolinium Oxysulfide Scintillator vs. the Plastic Scintillator**

Tables 2-2 and 2-3 contain the recorded signal and reference counts from the two dosimeters as well as the calculated output factors and net counts for field sizes of 3x3, 5x5, 10x10, and 15x15 cm. The net plastic counts decreased with increasing field size which was unexpected given the higher output factor. This led to the measured output factor decreasing with field size when normalized to a standard 10x10 field. The reverse effect was observed in the gadolinium dosimeter which signal fiber and reference fiber responded proportionally to changes in field size. The importance of these results indicate how well the signal counts correlate with the reference counts in the gadolinium scintillator over all field sizes, where with the plastic scintillator, the proportion begins to saturate after 10x10 cm significantly. The gadolinium dosimeter measures the output factor very well below 10x10 cm, but over under responds at field sizes greater than 10x10 cm.

Figure 2-4 illustrates the effect of the pulse repetition rate. Although the instantaneous dose rate isn’t changing, at higher frequencies counts from one pulse may start to overlap with counts from next pulse. The magnitude of this effect would have been expected from the gadolinium oxysulfide scintillator given its longer decay time. However, the plastic scintillator demonstrated higher repetition frequency dependence. Overall, the effect appears to be negligible given the magnitude of the variation in response. The overall trend appears to be linear with fractional losses equating to roughly 1.951 $\mu$s/MU and 5.686 $\mu$s/MU for the plastic and gadolinium
respectively. It should also be noted that the sensitivity between the two dosimeters is vast as the gadolinium was 30 times more sensitive to radiating despite a much lower detector mass and equivalent coupling efficiencies.

**Stem Count Behavior**

Data was acquired from various radiation sources (sources A [6MeV electron], B, and D) delivering variable quantities of radiation to the stem of the dosimeter by varying the field size. The scintillator, or sensitive element, was shielded to isolate the stem counts. The variable length was achieved by adjusting the field size on each of the sources and providing limited backscatter material to eliminate field effects. The counts from each source were normalized to the maximum field size reading to create a plot of normalized stem counts vs. field size. This representation is illustrated in Figure 2-5. The slope of the x-ray sources and proton source is comparable (0.038 relative counts/cm vs. relative 0.0329 counts/cm). Note that the protons energy is quantified in terms of the range of the particle. 14 cm proton is the nominal energy of clinical proton therapy with values ranging from 5 to 23 cm. The 1000 MU per min PRF was included to illustrate an observed effect when the PRF was set very high the counts would decrease, although this effect didn’t happen all of the time as in Figure 2-4.

**System Time Resolution**

**Continuous sources**

Radiation sources B, C, and D were used to evaluate the PSD dead time from different particle types and energies. The data was fitted by placing the measured inverse count rate data (y-axis) along with the inverse dose rate (x-axis) and fitting the data to a one-degree polynomial function. Figure 2-6 graphs the data collected from the polystyrene scintillator irradiated from sources B, C, and D. Calculated dead times from
Table 2-4 indicate that dead time is affected by the radiation source and energy of the ionizing particles. The measured dead time increases from the induction of Cerenkov radiation into the background fiber. It is also noteworthy that the slope of the fitted functions that corresponds to the sensitivity when corrected for dead time of the cobalt 60 source and the 14 cm proton source is 9.78 million counts per Gy and 10.1 million counts per Gy respectively. The nonparalyzable model is an excellent representation of the count rate data and very useful for calibrating the dosimeter for sources that are known to be energy independent for polystyrene.

**Pulsed x-ray source**

Dose rate effects were analyzed using the PSD system by varying the source to detector distance (SDD) to generate a series of dose rates for a 6MV photon beam. The measurements were preformed free in air using an ionization chamber with buildup cap to measure the charge collected at each position along the beams central axis. The PSD measured scintillating counts by subtracting the signal counts from the reference counts. Figure 2-7 A shows the dosimeter counts vs. charge collected by the ion chamber. The more charge collected indicates a higher dose rate since the dosimeter was positioned closer to the source. Charge greater than 6.7 nC resulted in a decrease which is not indicative of nonparalyzable dead-time behavior as shown in Figure 2-7 B. The dose rate cannot be quantified due to the unknown measurement time of both dosimeters, therefore the total charge was considered to be proportional to the measured dose rate. The counting times in both cases could be regarded as the same since the exposures were kept consistent at 100 MU at 400 MU/min. The y-intercept was calculated using the data that was lower than the threshold as shown in Figure 2-7 B. The intercept ($\alpha$) is shown in the last row of Table 2-5.
Percent Depth Dose Curves

Figure 2-8 compiles the four PDD curves from the 5x5, 10x10, 15x15, and 20x20 cm field sizes for the raw, corrected, and actual PDD data. The corrected PDD data were calculated by subtracting the background counts and using Equation 2-7 except instead of count rates, restricting $m$ to counts detected and the dead time $\tau$ to a correction factor $\alpha$ that accounts for the dead time divided by the counting time to find the count rate. This was done because without proper equipment it is possible to determine the average or instantaneous count rate. The dosimeter was exposed to 100 MU's of x-rays delivered at 400 MU/min (same as in the previous experiment) consistent throughout all measurements. The PDD curve was measured along the beams central axis at 100 SSD to the water phantom. The data was normalized to the 3-cm measurement location. The value of $\alpha$ used to correct the raw data is shown in Table 2-5.

Random Phenomena

When performing measurements at high photons and electron using the PSD system random spontaneous pulses were detected by PMT. These occurrences were random and independent of the detector. Figure 2-9 illustrates the effect of a 6MV photon beam performing PDD measurements from the previous experiment. On occasion, an isolated high count-rate pulse would be detected by one PMT but not the other. These events would appear frequently and the measurement would be discounted. Figure 2-10 is a count rate plot from two different experiments illustrating the sampled counts at 500 ms at increasing optical length exposed in the cobalt irradiator. After a certain count rate threshold the counts would decrease despite more optical in the irradiator. In both cases the counts in PMT’s would drop. These instances
demonstrate the unusual behavior of the detection system in a high energy radiation environment and may provide further insight into the experimental counting issues associated with the PSD detection system.

Discussion

The background subtraction is a proven and useful method for correcting the stem response in the radiological and proton therapy detection applications; however, is an inadequate method in radiotherapy CLINAC dosimetry using this PSD detection system. The effects of dead time along with utilizing these correction methods in radiological imaging dosimetry will be examined in the subsequent chapter. The other studies that used the background subtraction method experimented with PMT’s equipped or other photon detectors use charge integration or operate their detectors in current mode. Detectors that are applied to radiation dosimetry at high event rates are normally operated in current mode. The PMT system in this study contains an amplifier, comparator, pulse shaper, and counter system designed to digitize signal and transmit directly to a personal computer. The detector operates in pulse mode which has shown in this study to make it susceptible to pulse overlap and loss of information.

The nonparalyzable model in general, is an acceptable model for making proper corrections to obtain accurate dosimetry for this dosimetry system, however is limited in electron/x-ray radiotherapy by the instantaneous dose rate. Dead time corrections may play a future key role in proton therapy applications for dosimetry measurements in the Bragg peak region of the proton PDD curve. Publications have cited a linear energy transfer (LET) dependence that using this model may be partially correctable.81-83 The charged particle LET, often described in terms of the linear mass stopping power, can be corrected for ionization quenching by analyzing the energy deposition rate (dE/dt)
which can be quantified readily in counts per second by the detection system. Both quantities increase at lower proton kinetic energies however in varying magnitude since $dE/dx$ is proportional to one over the particle’s velocity squared and $dE/dt$ is proportional to one over the velocity.

Scintillation counting systems, especially the one investigated in this study, are limited in their ability to capture the entire signal at high count rates. This is because of dead-time effects that prohibit the system from distinguishing between pulses separated by a finite width. It is quite evident that the dead time of the plastic scintillator is increased when exposed to radiation capable of producing Cerenkov radiation. This is thought to be directly tied with the transient absorption effects happening within the optical fiber. The gadolinium dosimeter proved to be excellent at overcoming the field size effects producing accurate results in the smaller field sizes where scatter is less prevalent. The success of the gadolinium dosimeter was attributed to its long decay time small length, and emission spectrum. The dose rate experienced by these dosimeters from radiation delivered from a LINAC using the values from Table 2-1 is anywhere between 25 to 100 Gy/s (1,000 rads/s) which is very high when compared to the other clinical modalities. However the results from the cobalt experiment prove that even at lower dose rates the quenching or extended dead time is present.

The data by Mattern shows that delivery from pulsed radiation significantly increases the absorption of light photons within an optical fiber owing to a transient recovery of defects. These transient effects contribute to the dead time of the entire counting system by affecting the time that no scintillation light can be transmitted after the optical fiber is irradiated. Due to the short decay time of polystyrene, the fraction of
counts reaching the PMT was determined by the amount of luminescence created within the optical fiber. However, the pulsing of the radiation source clearly has a strong effect causing it to deviate from the nonparalyzable model at high dose rates and pulse rates. It also demonstrates a maximum count rate threshold that will cause even more losses if the count rate exceeds the maximum. This was consistent with the results presented by Daams. It still remains unclear why the stem count rate plays such significant role in increasing the dead time. Further analysis such as the scintillator time intensity profile and the emission spectrum of the scintillator under a diagnostic beam compared to radiotherapy may provide further insight.

**Conclusion**

This investigation has led to two very important results. First, the background subtraction method is an acceptable method for low-energy x-ray dosimetry for applications when Cerenkov production is not a concern. From this result, the correction factor may be found by the methodology discussed in this which is based on correlating a linear response to varying field sizes between the signal and reference fiber. This method is superior than just solely measuring a PMT correction factor as it accounts for all the parameters that contribute to the loss of signal. The second and probably most impactful conclusion is the effect of dead time in high energy scintillation counting experiments. However, in most of the cases it was correctable using the nonparalyzable model correction method.

When constructing the ideal PSD it should have a long temporal response and emit light with a similar wavelength as the gadolinium oxysulfide scintillator. It should also be characterized for dose rate dependence and not rely on the pulse repetition performance as a means of evaluating count rate effects since they are different.
Ideally, with the proper equipment these measurements could be potentially investigated to analyze the light spectrum of the gadolinium and PSD as well as profile the emitted pulse temporal response.
Table 2-1. Measured pulse width at half maximum and pulse period for the Varian Trilogy accelerator. 6 MV x rays, Pulse width = 3.90 μs.

<table>
<thead>
<tr>
<th>Average dose rate (MU/min)</th>
<th>Fraction of pulses</th>
<th>Period (ms)</th>
<th>Dose per pulse (Gy/pulse)</th>
</tr>
</thead>
<tbody>
<tr>
<td>600</td>
<td>1</td>
<td>2.77</td>
<td>2.78 x 10^{-4}</td>
</tr>
<tr>
<td>500</td>
<td>0.8</td>
<td>2.77</td>
<td>2.78 x 10^{-4}</td>
</tr>
<tr>
<td></td>
<td>0.2</td>
<td>5.55</td>
<td>2.78 x 10^{-4}</td>
</tr>
<tr>
<td>400</td>
<td>0.5</td>
<td>2.77</td>
<td>2.78 x 10^{-4}</td>
</tr>
<tr>
<td></td>
<td>0.5</td>
<td>5.55</td>
<td>2.78 x 10^{-4}</td>
</tr>
<tr>
<td>300</td>
<td>1</td>
<td>5.55</td>
<td>2.78 x 10^{-4}</td>
</tr>
<tr>
<td>200</td>
<td>1</td>
<td>8.33</td>
<td>2.78 x 10^{-4}</td>
</tr>
<tr>
<td>100</td>
<td>1</td>
<td>16.66</td>
<td>2.78 x 10^{-4}</td>
</tr>
</tbody>
</table>

Table 2-2. Plastic counts from signal and reference fibers as well as output factors normalized to a 10x10 cm field size from a 50 MU irradiation.

<table>
<thead>
<tr>
<th>Field Size (cm x cm)</th>
<th>Signal fiber (#)</th>
<th>Reference fiber (#)</th>
<th>Net Counts (#)</th>
<th>Output Factor (measured)</th>
<th>Output Factor (accepted)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3 x 3</td>
<td>48210</td>
<td>25444</td>
<td>22766</td>
<td>0.302</td>
<td>0.681</td>
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<tr>
<td>5 x 5</td>
<td>86481</td>
<td>47575</td>
<td>38906</td>
<td>0.517</td>
<td>0.793</td>
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<tr>
<td>10 x 10</td>
<td>178801</td>
<td>103512</td>
<td>75288</td>
<td>1.000</td>
<td>1.000</td>
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<td>15 x 15</td>
<td>180610</td>
<td>127688</td>
<td>52921</td>
<td>0.703</td>
<td>1.343</td>
</tr>
<tr>
<td>20 x 20</td>
<td>182008</td>
<td>136519</td>
<td>45489</td>
<td>0.604</td>
<td>1.477</td>
</tr>
</tbody>
</table>

Table 2-3. Gadolinium counts from signal and reference fibers as well as output factors normalized to a 10x10 cm field size from a 50 MU irradiation.

<table>
<thead>
<tr>
<th>Field Size (cm x cm)</th>
<th>Signal fiber (#)</th>
<th>Reference fiber (#)</th>
<th>Net Counts (#)</th>
<th>Output Factor (measured)</th>
<th>Output Factor (accepted)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3 x 3</td>
<td>1551329</td>
<td>155107</td>
<td>1380711</td>
<td>0.691</td>
<td>0.681</td>
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<tr>
<td>5 x 5</td>
<td>1785994</td>
<td>201427</td>
<td>1564424</td>
<td>0.783</td>
<td>0.793</td>
</tr>
<tr>
<td>10 x 10</td>
<td>2288194</td>
<td>263002</td>
<td>1998892</td>
<td>1.000</td>
<td>1.000</td>
</tr>
<tr>
<td>15 x 15</td>
<td>2684579</td>
<td>301682</td>
<td>2352729</td>
<td>1.177</td>
<td>1.343</td>
</tr>
<tr>
<td>20 x 20</td>
<td>2997395</td>
<td>331994</td>
<td>2632202</td>
<td>1.317</td>
<td>1.477</td>
</tr>
</tbody>
</table>
Table 2-4. Calculated dead times from various radiation sources

<table>
<thead>
<tr>
<th>Scintillator</th>
<th>Radiation source</th>
<th>Dead time parameter $[\tau]$ (ns)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gadolinium</td>
<td>Orthovoltage</td>
<td>82.3</td>
</tr>
<tr>
<td>Gadolinium</td>
<td>Cobalt-60</td>
<td>698</td>
</tr>
<tr>
<td>Polystyrene</td>
<td>Orthovoltage</td>
<td>629</td>
</tr>
<tr>
<td>Polystyrene</td>
<td>Proton</td>
<td>685</td>
</tr>
<tr>
<td>Polystyrene</td>
<td>Cobalt-60</td>
<td>5356</td>
</tr>
</tbody>
</table>

Table 2-5. Calculated dead time parameters of a 6 MV photon beam free-in-air and percent depth dose (PDD) measurements.

<table>
<thead>
<tr>
<th>Measurement technique</th>
<th>Dead time parameter $[\alpha]$ (unitless)</th>
</tr>
</thead>
<tbody>
<tr>
<td>PDD field size: 5 x 5</td>
<td>3.10E-05</td>
</tr>
<tr>
<td>PDD field size: 10 x 10</td>
<td>1.01E-05</td>
</tr>
<tr>
<td>PDD field size: 20 x 20</td>
<td>1.13E-05</td>
</tr>
<tr>
<td>PDD field size: 20 x 20</td>
<td>1.10E-05</td>
</tr>
<tr>
<td>Free-in-air</td>
<td>1.05E-05</td>
</tr>
</tbody>
</table>
Figure 2-1. Schematic illustrating the acceptance cone of the optical fiber in relation to the Cerenkov emission cone of the electron transversing through the fiber.

Figure 2-2. Pulsing kinetics of a typical LINAC. X-rays and electrons are delivered in a series of pulses which are actually a set of smaller pulses of radiation released in sync with the magnetron.
Figure 2-3. PMT time response from a finite light pulse at the photocathode.
Figure 2-4. Pulse repetition rate effects on the dosimeter’s response. PRF values correspond to pulse frequencies ranging from 2.77 ms up to 16 ms. These frequencies are represented graphically as 100 MU/min to 650 MU/min.
Figure 2-5. Optical fiber response to varying lengths of radiation exposures delivered in a series of field sizes to the optical fiber. A linear fit was applied to the data to analyze the linearity of the stem signal generated in the background fiber. Error bars are shown to represent 1 standard deviation from the mean.
Figure 2-6. Plot of inverse count rate and dose rate to measure dead times from cobalt 60, orthovoltage (350 kVp x-rays), and clinical protons (12-cm range in water) used in proton radiotherapy irradiations. Data fitted to a linear model to calculate the y-intercept. Dead time values are listed in Table 2-4. Error bars are included and represent 1 standard deviation from the mean.
Figure 2-7. Megavoltage beam irradiation free-in-air. Top figure (a) shows the net dosimeter counts from different dose rates. The bottom figure (b) is those data arranged to fit to the nonparalyzable dead time calculation. The dose rates were obtained by moving both the dosimeter and the ionization chamber along the gantry axis.
Figure 2-8. Percent depth dose curves for field sizes 5x5 cm (a) 10x10 cm (b) 15x15 cm (c) and 20x20 cm (d) containing the measured curves using the ion chamber, PSD raw data, and PSD corrected data. Red markers indicate the raw data, green the corrected data, and blue is the measured data using an ionization chamber. PDD was normalized to the 3-cm measurement depth.
Figure 2-9. Spontaneous pulse from the detection system within the signal optical fiber but not present in the reference optical fiber. Each data point corresponds to the counts sampled by each PMT at 50 ms.
Figure 2-10. Counting profiles from incremental insertion of the PSD into the cobalt irradiator. Both counting profiles (blue and green) are captured from various amounts of optical fiber irradiated. Light color line is the signal optical fiber and the dark color line is the reference fiber. After the maximum count rate (normalized to 1) the counts began decrease despite more detector exposure.
CHAPTER 3
DEVELOPMENT OF AN ENERGY INDEPENDENT FIBER-OPTIC DOSIMETER FOR
RADIOLOGICAL DOSIMETRY

Energy Dependence in Radiological Dosimetry

The term energy dependence is used to describe the inaccuracy of using a dosimeter, caused by the energy of the incident radiation, to measure absorbed dose from ionizing radiation. In the low-energy range of medical x-rays, it is often attributed to the difference between the material composition of the dosimeter and the evaluated biological tissue. Variations in density, atomic weight, and effective atomic number lead to differences in photon-energy absorption resulting from more or less photon interactions which deposit energy inside the sensitive element. Diagnostic energy x-rays deposit a majority of their energy through photoelectric absorption making the response of the dosimeter highly dependent on the effective atomic number ($Z_{\text{eff}}$) of the sensitive material. The mass-energy absorption cross section ($\mu\text{en}/\rho$), defines the probability of photon interaction that results in energy transfer to localized charged particles. Since the coefficient is inherently energy dependent, the effective mass-energy absorption cross section is calculated by weighting each energy component based on the incident photon energy spectrum. Further complicating the issue, these coefficients change drastically with respect to energy in the diagnostic energy range. It is extremely difficult to directly measure the ratio of mass energy absorption coefficients; which is why dosimeters need to be calibrated under their measurement conditions. The reference dosimeter used for calibrating is an ionization chamber, which is known to be the gold standard in radiological dosimetry. Diagnostic ionization chambers show a variation in energy response of 5% from 1 keV up to 1.33 MeV. They produce the most reliable measurements, making them ideal candidate to calibrate other types of
dosimeters. However, they are used primarily for machine calibration and quality assurance. Table 3-1 taken from the International Atomic Energy Agency (IAEA)\textsuperscript{85}, lists the recommendation specifications of a reference class for various imaging modalities. The disadvantage of ionization chambers is their size which makes them unsuitable for measurements in phantoms that do not already accommodate these chambers.

In order to perform patient dosimetry using phantoms, it would be ideal to use a dosimeter that is small and compact, as well energy independent, all while preserving the other characteristics that are desired in radiation dosimetry. Currently there exist a select group of solid state dosimeters that have been characterized for clinical radiological dosimetry.\textsuperscript{86-88} These are: thermoluminescent dosimeters (TLDs) optically stimulated luminescent dosimeters (OSLs), and metal-oxide-semiconductor field-effect transistors (MOSFETs). The first two are crystal structures able to retain a fraction of the absorbed energy within metastable states that result from defects in their material. A crystallized lattice structure, stores energy in trapped electrons that maintain this metastable state until they are excited enabling them to recombine with neighboring electrons and release energy in the form of light or heat. The amount of light emitted and collected is proportional to the amount of energy absorbed. The two methods of exciting the trapped electrons within the crystal lattice structure are in the form of heat, thermoluminescence, and light, optically stimulated luminescence. The dual-MOSFET dosimeter is a type of p-n junction semiconductor that is two diodes on the same chip operating at two different gate biases. This architecture produces more signal and allows for real time monitoring. More stability is gained and minimal temperature/pressure corrections are required.
**Thermoluminescent Dosimeters**

TLDs are an acceptable dosimeter for performing patient dosimetry. A review of using TLDs for dosimetry in diagnostic radiology was done by Zoetelief et al.\(^8^9\). Thermoluminescent crystals are very sensitive to radiation making them exceptional low-level radiation detectors. They are especially effective when fabricated in the form of loose powders, disks, squares, and rods. Only a small amount of material is needed to form the sensitive region. They require an extensive amount of preparation and handling to avoid fading and annealing processes that affect measurement. For low-energy dosimetry the sensitive element material is an inorganic compound, Lithium Fluoride (LiF), with Mg, Cu, P, Ti activators. In particular LiF:Mg:Ti (TLD-100) has been commonly employed in clinical dosimetry because of its near tissue-equivalent effective atomic number, 8.2, compared to 7.4 for soft tissue. LiF:Mg:Cu:P (GR-200) is another type of TLD material that offers 30 to 50 times more sensitivity and exhibits less supralinear behavior than Td-100, although has demonstrated a more significant energy response to low energy x rays. Response in sensitivity for LiF:Mg:Ti has been characterized within 9% over diagnostic x-rays ranging from beam qualities of 1.48 to 6.33 mm Al.\(^8^8\) TLDs are not ideal for clinical diagnostic x-ray beams due to an extensive annealing process which de-sensitizes the crystals and a non-linear dose response (supralinearity) over diagnostic energies.

**Optically Stimulated Luminescent Dosimeters**

Optically stimulated luminescence dosimeters (OSLDs) offer excellent sensitivity to radiation as well as are manufactured with a small size making them very useful for in-phantom dosimetry. Over the past few years, they have started to replace TLDs as the most popular instrument for radiological dosimetry because of superior
The sensitive element of a NanoDot OSL dosimeter is Al$_2$O$_3$, effective Z of 11.3, which causes an over-response to low energy photons as a result of high contribution from the photoelectric effect. The sensitive volume consists of a cylinder 0.5mm in diameter that extends to 0.1 mm in length. As a result of their small size good spatial resolution is obtained. OSLs are calibrated to measure point doses using a small volume ion chamber. TLD and OSLD light output is measured using a light photon counting device such as a photomultiplier tube (PMT). Commercially available dosimetry devices employ OSL technology to create small compact dosimeters for a variety of dosimetry applications. In terms of energy dependence, 15% variation was observed by one study over CT x-ray beams of 80 and 120 kVP. Another study evaluated the energy dependence over the entire diagnostic range and a variation of 33% across beam spectrums with effective energies from 25 to 45 keV.

**Metal-Oxide-Semiconductor Field-Effect Transistor**

Like the dosimeters described above, MOSFETs have a relatively small size, which aids in their ease of use. They produce instantaneous measurements making them a lot faster than the OSLD and TLD. The limitation of using MOSFETs is that they do not last as long/are not as reusable as other dosimeters leading to a long term cost due to progressive loss in sensitivity. MOSFET dosimeters have demonstrated undesirable reproducibility fluctuations that gradually occur at low keV energies over long time periods. This requires MOSFETs to be recalibrated after a certain amount of usage. Their energy correction factor has shown to approximately double going from a 30 kV x-ray source to a 150 kV source. Studies have concluded that the under-response in lower keV energies, MOSFETs may be the least ideal for low-energy photon dosimetry.
Tissue Equivalent Fiber-Optic Dosimeter

Background

Plastic scintillating dosimeters (PSDs) are a promising dosimetry tool for low-energy photon dose measurements. They have been strictly utilized in research-based studies, and have yet to enter into the commercial marketplace. PSDs are highly sensitive, exhibit good counting statistics, and demonstrate approximate tissue equivalence in the diagnostic energy range (less than 150 keV). PSDs produce instantaneous measurements allowing for real-time dosimetry and require no post-irradiation processing. Ideally, the response of a PSD relative to soft tissue would remain independent of energy, direction, and dose rate. Studies that have characterized PSDs have observed them to be susceptible to these effects. Angular dependence is attributed to misalignment between the sensitive element and the optical waveguide. However, from a dosimetry perspective, placing the dosimeter in a scattering medium significantly minimizes this effect. The most concerning issue of using PSDs for medical x-ray imaging dosimetry is relating the response of the dosimeter to the difference in energy absorption properties between biological tissues and plastic. Typically PSDs used for radiological exposures are calibrated to the energy of the x-ray spectrum to mimic the response of air-filled gas detectors which have been shown to resemble soft tissue. The energy dependence of the PSD is also manifest as a variation in sensitivity as a function of depth in tissue equivalent material that has been characterized as high as 10% across 16cm of solid water for a 120 kV beam. Variation in sensitivity has been attributed to beam quality changes induced by beam hardening and scatter. Lessard et al. developed a systematic approach for obtaining correction factors relating the absorbed energy per unit mass of polystyrene to water to
offset the energy response of polystyrene for x-ray beams characterized with an effective energy less than 86 keV. For the diagnostic x-ray sources examined, it was demonstrated that the correction factor was directly correlated to the spectrally averaged mass energy-absorption coefficients of polystyrene and water demonstrating the application of large cavity theory to radiological dosimetry. Studies that have investigated other dosimeters, with less air equivalence than plastic scintillators, have noted that the response at various depths have shown good agreement with ionization chambers. The similarity in response despite differences in material composition of each dosimeter’s sensitive volume may prove depth-dependent correction factors are nonvarying due to little change in the incident x-ray spectrum. This evidence suggests the role of quenching in plastic scintillators, which has been characterized below 125 keV, may play a larger role in depth-dose measurements due to lower energy contributions from scattered photons. The effect of quenching was extensively studied by Frelin who investigated the response of plastic scintillators to low energy photons and electrons, and observing the linearity of plastic scintillators deviated under 200 keV for photons and 100 keV for electrons.

**Two Scintillators Compacted into a Single Dosimeter Package**

Hyer et al. and Moloney have characterized polystyrene and gadolinium oxysulfide (GOS) fiber-optic dosimeters respectively for radiological dosimetry. In their respective investigations they have observed that both dosimeters demonstrate positive energy dependence with respect to tube potential. A correlation from the ratio of the response between the two dosimeters would prove to be extremely useful for correcting the depth dependence of the PSD, especially due to the sensitivity of GOS scintillator to scatter radiation. In theory, changes in sensitivity of the ratio of plastic counts to
gadolinium counts with depth would directly correlate with the change in sensitivity of the PSD. This manuscript characterizes the effectiveness of using a dual scintillator dosimeter to resolve the energy dependence of the PSD. Due to the high sensitivity of GOS to x-rays, the scintillator needs to be evaluated for dead time, in a similar fashion explained in Chapter 1, however using a radiographic x-ray tube. Table 3-1 tabulates the physical properties between both scintillators including the effective atomic number, decay time, and rise time. Note the differences in decay time between the two scintillators.

**Materials and Methods**

**Construction of the Dosimeter**

The dosimeter was constructed as a package containing three cylindrical 500-micron diameter multi-mode optical fibers (Eska CK-20, Mitsubishi Rayon Co, Tokyo, Japan), two of which were coupled to scintillators, with the remaining fiber left uncoupled to quantify radiation induced fluorescence as described in Chapter 2. This reference fiber was positioned adjacent and parallel to other fibers to quantify all stem counts originating in the scintillator waveguide. These stem counts were subtracted from the measured signal in the optical fibers.

The polystyrene scintillating element (BCF-12, St Gobain Crystals) was manufactured as a cylindrical fiber of similar dimension (500 microns) to the optical fiber. The coupling surface of each fiber was polished using the technique outlined in Chapter 1. The scintillator was optically coupled to the optical fiber using a transparent adhesive gel (Krazy Glue®) and 1 cm of 0.12 cm heat-shrink tubing (Fisher Scientific) to create tighter coupling. After coupling, the opposite end of the plastic scintillator was cut to the desired length of 3mm and coated with the reflective coating.
The GOS scintillator was obtained in the form of a 100μm layer from a rectangular sheet of an image intensifying screen for clinical screen film mammography (Kodak MIN-R 2 Cassette, Eastman Kodak Company). The GOS was coupled to a polished optical fiber by applying a thin coat of the adhesive gel to the polished end of the optical waveguide and placing the fiber onto the slab while exerting pressure to enhance the binding process. After a short time, the optical fiber was removed along with the attached thin layer of GOS coupled to the end of the fiber. To provide structural support and shielding from ambient light, each scintillator-fiber element and the reference fiber were encased in heat shrink tubing to create a single package containing all three fibers.

A single PMT is required for monitoring each scintillator element. Therefore, the fibers were branched off from one another to allow for an individual connection with the PMT array. The uncoupled end of each optical fiber was fitted with a female SMA 510 connector (SMA-905) using a SMA epoxy (FiberFin, 1656 resin, #80 hardener). The remaining length of uncovered optical fiber was wrapped with heat shrink tubing. A male SMA optical fiber adapter (E5776-51, Hamamatsu Corporation, Bridgewater, NJ) for each PMT was integrated within the protective PMT housing unit to interface each fiber to the PMT. Maintaining a small-size of the dosimeter is important for dosimetry measurements involving anthropomorphic phantoms, since the size is an important factor in how easily the dosimeter can be integrated with the phantom. Figure 3-1 illustrates the size of a completed dosimeter relative to a U.S. quarter.

The photon detection system, used by Hyer, was designed for low-level photon counting, using an array of PMTs each mounted with a microcontroller that permits data
transfer. Counting data was buffered and transferred through a hub to a computer for readout. A USB/serial hub ( UPORT 1610-8, Moxa Inc, Brea, CA) provided a pathway to transmit data between the computer and the PMT array. Data acquisition settings and communication between the computer and PMTs were controlled by a custom MATLAB (MathWorks, Natick, MA) graphical user interface program permitting the user to monitor counting data, PMT gate time, as well as graph and save all acquired data.

**Determining the System Time Resolution and Scintillation Efficiency**

The time resolution, or dead time of the dosimeter, was evaluated by performing a series of measurements with a variable exposure rate to the scintillator, which was achieved by varying the x-ray tube current. To measure the true count rate, a clinical CT thimble chamber (Radcal® RC0.6CT, Monrovia, CA) was used to quantify the exposure rate and exposure time. This ionization chamber was described in AAPM Report No. 111 as an appropriate instrument for CT dosimetry. The observed count rate and exposure rate were calculated by averaging the net scintillation counts or exposure and dividing by the exposure time. For this method, the true count rate is expressed in terms of the exposure rate, as the two are related by a conversion factor.

The dosimeter dead time was calculated by expressing the data in terms of the inverse observed count and the inverse exposure rate stated in Equation 2-9. The dead time can be solved for using linear regression to find the optimal y-intercept of the inverse of the true count rate and observed count rate data:

\[
\frac{1}{m} = \frac{1}{\varepsilon D} + \tau
\]  

(3-1)

In Equation 3-1, the theoretical value of \( n \) is the product of the exposure rate and a conversion factor \( \varepsilon \) relating the number of true detected counts to the measured
exposure/ionization rate. The value of $\varepsilon$ is the ratio of the detected count rate to the exposure rate. The two are related by the absorbed dose rate to the scintillator:

$$\dot{D}_S = \dot{X} \frac{W}{\varepsilon} \left( \frac{\mu_{en}/\rho}{\mu_{en}/\rho}_{\text{AIR}} \right)^S \frac{S_{\text{sys}} n}{m_{\text{det}}}$$  \hspace{1cm} (3-2)

Equation 3-2 is valid under charged-particle equilibrium and the condition that the Roentgen is measurable; otherwise, the expression is invalid since the relationship between dose and exposure breaks down. The 0.51mm heat shrink tubing that encases the scintillator provides sufficient buildup to achieve charged-particle equilibrium for diagnostic beams as the CSDA range of a 150 keV electron in polystyrene is 0.3 mm.$^{106}$ $S_{\text{sys}}$ is defined as the energy absorbed by the scintillator per detected light photon and is further examined by Beddar et al.$^{107}$ The term $m_{\text{det}}$ is the mass of scintillator material in the dosimeter. Combining terms the theoretical definition of the conversion factor relating the true count rate and the dose rate may be written:

$$\varepsilon = \left( \frac{\mu_{en}/\rho}{\mu_{en}/\rho}_{\text{AIR}} \right)^S \frac{m_{\text{det}}}{S_{\text{sys}}}$$  \hspace{1cm} (3-3)

Equation 3-3 provides the conversion factor between the true count rate and measured dose rate in air. To calculate $\tau$, the absolute value of $n$ is unnecessary, since the y-intercept solution is dependent only on the relative value of $n$. Therefore when calculating $\tau$, $n$ may be expressed in terms of exposure rate and as long as the energy spectrum remains unchanged, the conversion between the true counts and the ion chamber reading remains constant. The term $\varepsilon$ may be derived from fitting the data to Equation 3-1. With the use of Monte Carlo techniques, the spectrally averaged mass energy-absorption coefficients can be calculated to provide a value for the energy to light conversion efficiency per unit mass for the investigated energies. From this
information the variation in $S_{sys/m}$ may be evaluated for each investigated tube potential.

**Exposure Measurements**

**Free-in-air measurements**

A CPI Indico Plus RF high frequency x-ray tube (Communications & Power Industries International Inc., Georgetown, Ontario, Canada) with a variable tube current output was used to deliver a series of different radiological x-ray spectrums. The dosimeter was positioned 50.8 cm from the tungsten target to maximize the incident exposure rate. The x-ray tube current was varied to produce a range of exposure rates irradiating the dosimeter (25-500 mA), while maintaining the tube current time-product (25 mAs), field size (17 cm x 17 cm), and beam energy. The exposure rate was measured using the ionization chamber placed adjacent to the dosimeter. At each exposure, the net scintillating counts, exposure, and exposure time were recorded. Tube potentials of 140, 120, 100, 80, and 60 keV were used to investigate the effect of photon energy on the time response characteristics of the detection system.

**CTDI phantom measurements**

The dosimeter was exposed under varying layers of Polymethyl methacrylate (PMMA) by using multiple hole locations (1.66, 9.67, 16.0, 22.35, and 30.35 cm) on a PMMA CTDI head phantom nested within a body phantom (Radcal®, TO CTDI, Monrovia, CA). Using a clinical Toshiba Aquilion ONE volumetric CT scanner (Toshiba America Medical Systems Inc., Tustin, CA) radiation was delivered with the tube parked at the 12 o’clock position. Depth-dose measurements were acquired at all four tube voltages: 80, 100, 120, and 135 kVp using the large focal spot and medium bowtie filter at 200 mAs tube current time-product for 1 second with a collimation thickness of 16 cm.
The surface of the phantom was positioned 40 cm from the focal spot. The ionization chamber was used to record reference air-kerma measurements at each hole location. At each measurement location, the center of the ionization chamber and dosimeter were centered with the central axis of the fan beam. Five measurements were recorded at each location and all tube potentials to generate a calibration curve that would be used to correct the plastic scintillator reading as a function of depth for each tube voltage.

**Angular dependence**

Due to the architecture of the dosimeter there is lack of symmetry within the sensitive element from the orientation of the plastic and gadolinium scintillators. As a consequence, the angular direction of the radiation may affect the response ratio from perturbation of the x-ray beam. Therefore the angular dependence of the response ratio of the gadolinium signal to the plastic was evaluated using the CT scanner which provides fixed beam angle exposures. Static beam angles delivered exposures at 120 kVp and 100 mAs at the dosimeter which positioned at the machine isocenter free-in-air with the table removed from the beam. Two sets one measurements were where the gantry moved one full rotation around the axis of the dosimeter and irradiated in 45° intervals, and another sweep where the dosimeter was rotated 90° and the gantry irradiated the dosimeter normal-to-axial. In this case 0° corresponded to x-ray hitting the tip of the dosimeter unperturbed, and at 180° incident from the back end of the dosimeter and attenuated by the optical fiber. Measurements were again acquired in 45° intervals.
Calculation of error

Performing linear regression using the least squares method to determine the system dead time and fiber sensitivity will induce error and produce an uncertainty associated with each calculation. This will be especially important when testing the dead time of the polystyrene scintillator, which is less sensitive to radiation, making the calculation more uncertain. When performing the regression analysis, three measurements at each of the six tube currents were taken into account, meaning eighteen data points were included in each dead time calculation. Equations 3-4 and 3-5 were used to find the error associated with each parameter of the straight line curve fit characterized by the slope and y-intercept.

\[ \sigma^2(a) = \frac{\sum \frac{1}{\sigma_i^2} \sum x_i^2 - \left( \sum \frac{x_i}{\sigma_i} \right)^2}{\sum \frac{1}{\sigma_i^2} \sum \frac{x_i^2}{\sigma_i^2} - \left( \sum \frac{x_i}{\sigma_i} \right)^2} \]  

\[ \sigma^2(b) = \frac{\sum \frac{x_i^2}{\sigma_i^2}}{\sum \frac{1}{\sigma_i^2} \sum \frac{x_i^2}{\sigma_i^2} - \left( \sum \frac{x_i}{\sigma_i} \right)^2} \]  

Equation 3-5 was used to find the error in calculating the dead time. Standard deviation (\( \sigma_i \)) was found by using both error propagation and by modeling the error of each background and signal measurement in accordance with a Poisson distribution. Measurement uncertainties from the ion chamber owing to uncertainty from the standard calibration factor were (0.7%), effect of beam-quality difference between calibration and measurement (2.0%), stem effect (1.0%), and dose rate dependence (2.0%). Considering all of these errors and error from the timer, the uncertainty in any
given exposure rate measurement was conservatively estimated using the TG-61 protocol as 3.5%.\textsuperscript{106}

The error in generating relative fluence energy spectrums from the Monte Carlo were evaluated by using the simulated statistical error calculated for each energy bin and propagating the error to generate an alternative relative energy spectrum. This spectrum is an equivalent spectrum representing the extremes possibilities of the detected energy spectrum. Using this alternative relative energy spectrum the uncertainty in the spectral weighted mass energy-absorption coefficients and quenching factors were recalculated to obtain a width of the uncertainty from the Monte Carlo simulations.

\textbf{Monte Carlo Methods}

\textbf{Generating photon and electron energy fluence data}

Each scintillator’s energy response from x-ray absorption properties was evaluated by simulating the initial photon energy spectrums using Monte Carlo techniques along with stopping power and energy absorption coefficients from the NIST database.\textsuperscript{108, 109} A source model of each x-ray tube energy output was created as a custom source subroutine within a general-purpose Monte Carlo radiation transport code, MCNPX version 2.7 (Los Alamos National Laboratory, Los Alamos, New Mexico). The required probabilistic information regarding beam energy, the shape of the CT x-ray fan beam due to the bow tie filter, and x-ray field collimation were all experimentally derived from measurements on each machine. The source subroutine for the Toshiba Aquilion ONE was created using the spectrum generator software with additional information to account the shape of the x-ray fan beam due to the bowtie filter and z-axis collimation. Methodologies presented in research conducted by Turner et al.\textsuperscript{110}
and Monica Ghita\textsuperscript{111} at the University of Florida were used in concert with a custom MATLAB\textsuperscript{™} script function and spectrum generator program to create the source subroutine in FORTRAN 90 (International Business Machine Corporation, Armonk, New York).

A tool for computationally modeling tungsten anode x-ray generators was developed by Siewerdsen based on the TASMIP model of Boone and Seibert.\textsuperscript{112, 113} The software, SPEKTR, allows user to generate tungsten anode spectrums for different filtrations across a broad range of kVp, filter material, and filter thickness. The equivalent x-ray spectrum of the CPI Indico Plus RF high frequency x-ray tube was created from three input measurement parameters: peak tube potential, and first and second half-value layer. With the software, an initial unfiltered tungsten spectrum is generated. This initial spectrum is iteratively transmitted through different thicknesses of various inherent filtration materials (lead, aluminum, copper, tungsten, and carbon) in 1 keV energy bins using attenuation coefficients from the NIST database for aluminum, graphite, lead, titanium, and copper. The simulated first half-value layer and second half-value layer were calculated at each iteration until the simulated first half-value layer is equivalent to the measured for each material. The simulated equivalent spectrum was chosen from the simulated second half-value layer spectrums of different filtrations based on the second half-value layer that most closely matched the measured second half-value layer.

With the creation of each source subroutine, the x-ray and electron spectra transported through each scintillator can be simulated by modeling the physical experiment. The convenience of generating a normalized energy spectrum is that the
quantity of photons generated is not necessary. Spectrum information was captured by binning F4 tallies in energy bins from 1 to 143 keV for photons. Electron energy tallies were compiled from 1 to 140 keV in 1 keV increments above 10 keV, 0.5 keV increments between 5 to 10 keV and 0.25 keV bins down to 1 keV. Higher resolution energy bins were required in the lower energy levels to account for the large changes in stopping power with energy. For example the difference in stopping power between a 2.25 keV and 2.5 keV electron is equivalent to a 40 keV and 140 keV electron.

The custom CT source subroutine was further validated with in-scanner CTDI measurements by Long at the University of Florida. First, a free-in-air measurement was made in the scanner to normalize photon output per mAs, Next, using a RadCal® 10-cm ion chamber, dose measurements were made at the center and peripheral hole locations of a CTDI head and body phantom. These dose measurements were than computationally simulated in MCNPX for output comparison yielding an average error less than 3%. The x-ray spectrum of both scintillators was validated by matching both first and second half-value layer measurements.

**Evaluation of ionization quenching from Birks’ model**

Once the electron spectrum is obtained, the magnitude of quenching experienced by the dosimeter may be determined. With the assumption that the electron fluence throughout the scintillator will remain unperturbed under electron equilibrium and the range of each electron is less than the diameter of the scintillator, each electron of energy $E$ can be modeled to deposit all of its energy into the scintillator. Therefore, the light output for a scintillator with scintillation efficiency $S$ from a photoelectron with initial kinetic energy $E$ within the sensitive volume may be expressed using the empirical unimolecular quenching model proposed by Birks and Brooks$^{114}$:
\[ L(E) = \int \frac{dL}{dE} = S \int_0^E \frac{1}{1 + kB \left( \frac{dE}{dx} \right)} dE \]  

(3-6)

The energy dependent quantity \( L(E) \), was evaluated at each energy using numerical integration over the entire kinetic energy of the electron. Stopping power information was obtained from the NIST ESTAR database, which tabulates electron and positron stopping power data. Published values for the quenching parameter \( kB \) for plastic scintillators range between 0.009 to 0.016 cm\(^{-2}\)MeV\(^{-1}\)g.\(^{104,115-117}\) A search of the literature found no reported quenching factors for GOS. Most inorganic scintillators that have been evaluated, parameterized the product of \( kB \) from Equation 3-6 between 0.008 and 0.001.\(^{76,117}\) The total quenching factor \( Q \) was calculated for each energy spectrum by integrating \( L(E) \) over the simulated electron energy fluence spectrum and dividing by the integral of the electron energy fluence and the scintillation efficiency product expressed in Equation 3-7.

\[
Q = \frac{\int_{0}^{E_{\text{max}}} \varphi(E') S \left( \int_{0}^{E'} \frac{1}{1 + kB \left( \frac{dE}{dx} \right)} dE' \right) dE'}{\int_{0}^{E_{\text{max}}} S \varphi(E') dE'}
\]  

(3-7)

The term \( S \) is energy independent and can be factored out of both integrals. All integrals can be evaluated discretely using the initial electron energy fluence data, quenching factor derived from Birk’s formula, and stopping power data. In Equation 3-7, \( E' \) is the secondary electron spectrum, and \( E \) is the kinetic energy of the electron as it losses energy within the scintillator.

Electron energy spectra for each scintillator were collected in energy bins from Monte Carlo surface tallies. Electron energy tallies were compiled from 1 to 140 keV in
1 keV increments above 10 keV, 0.5 keV increments between 5 to 10 keV and 0.25 keV bins down to 1 keV. The range of the photoelectrons was found to be less than the diameter of the scintillation volume. The CSDA range for a 125 keV electron is less than 40% the diameter of the plastic scintillator and 1/10\textsuperscript{th} the diameter of the GOS.

**Results**

**Count Rate Performance**

Light photons originating from the optical fiber and detected by the PMT were measured by placing a reference optical fiber next to the optical fiber carrying the signal from the scintillator. The counts from the reference fiber were correlated to counts in the signal fiber by placing the pair of optical fibers under the x-ray field and placing the scintillator outside the x-ray field and then irradiating different lengths of the stem at multiple beam qualities with tube potentials ranging between 60 to 140 keV. The data was used to generate a correction factor that relates the stem counts detected in the reference optical fiber to the stem counts detected in the signal fiber. Each scintillator demonstrated a linear response between detected counts in the background fiber and the scintillator fiber. This method was thought to be superior to taking measurements with the reference fiber occupying each PMT, because the correction factor could be dependent on parameters other than the PMT efficiency as discussed by Beddar et al.\textsuperscript{107}

The measured count rate data (y-axis) along with the air-kerma rate (x-axis) are plotted for the GOS scintillator and plastic scintillator in Figure 3-2 for the 120-kVp x-ray spectrum. These graphs show the response of both scintillating elements to varying dose rate. Figure 3-2 also illustrates the first degree polynomial fit to the data when presented in the form of Equation 3-1 to calculate the dead time ($\tau$) and the air kerma
sensitivity factor (ε) expressed in counts per air kerma. The values of τ and ε from the best fit to Equation 3-1 are tabulated in the first and second columns of Table II and Table III.

**Monte Carlo**

The computed values for the spectrally averaged mass energy-absorption coefficients between the scintillation and ionization chamber are tabulated in Tables 3-2 and 3-3 in the fourth column. These ratios were calculated using the normalized x-ray spectrum and the mass-energy absorption coefficients from the NIST database. The scintillation efficiency for each tube potential was calculated to examine the accuracy of energy correction factors based on the large cavity theory proposed by Lessard and point out any quenching effects from free-in-air measurements. In the columns to the right, the quenching factors calculated from Equation 3-7 using the highest and lowest values of the quenching parameter, kB, from the literature. The uncertainty in the gadolinium quenching factor calculation was negligible, while the plastic scintillator was less than half of a percent. The reported statistics by the Monte Carlo run were no more than 0.2% for gadolinium and 1.6% for polystyrene. It should be noted that the absolute scintillation efficiency was not able to be determined due to the difficulty in measuring parameters such as coupling efficiencies and transmission losses from the fiber cable.

The electron energy spectrum for 120 kV at various positions in the phantom is plotted in Figure 3-3. The average energy of the spectrum decreases at deep depths due to attenuation of the primary beam and energy contributed from low-energy scattered photons. When calculating quenching factors, GOS showed no change in quenching with depth, while polystyrene showed a slight deviation between 8 and 16 cm but a significant decrease after 16 cm.
The photon energy-absorption characteristics of both GOS and polystyrene relative to air maintained less than 5% variance between all locations for a 120-kVp x-ray beam, with polystyrene showing a lesser energy dependence. These results also show that the ratio of the mass energy-absorption coefficients between GOS and plastic shouldn’t vary by more than 6% across the entire phantom. Due to the small size of the gadolinium scintillator the x-ray energy spectrum at the deepest-hole location was statistically insignificant because of a low number of detected photons in each bin. The effect on the plastic scintillator from beam hardening is in fact minimized by the characteristic x-ray contribution of the GOS, which negates the higher energy characteristic x-rays of the tungsten anode. These characteristic x-rays dominate the energy spectrum of the GOS scintillator leading to a convenient slowly varying response as a function of depth. Figure 3-4 illustrates the relative x-ray energy fluence spectrums through each scintillator at the phantom top and center-hole locations.

**Energy Response**

Figure 3-5 plots the normalized sensitivity for each scintillator in terms of counts per Gy as a function of depth into the CTDI phantom for three different tube potentials: 100, 120, and 135 kVp. The data were normalized to the 120 kVp data point at the center-hole location. Each plot includes three sets of data: the GOS corrected for dead time, GOS uncorrected for dead time, and the plastic scintillator uncorrected for dead time. A dead time of 48 ns was used to correct for count losses. This 48 ns value was calculated by using the data from all energies and applying the 1st order polynomial best fit to find the dead time. The data were also examined to find the optimal dead time that would minimize the variance in sensitivity, which was found to be between 50 to 52 ns.
The decrease in sensitivity at greater depths was attributed quenching from the presence of low-energy electrons as seen in Figure 3-4.

The energy dependence of a dosimeter is often attributed to the variance in sensitivity of the dosimeter from the calibration condition. Plastic scintillator response factors are expressed in the quantity net counts per measured air kerma. To determine a correction factor methodology based on a given calibration condition an empirical relationship was evaluated to include all CT energies, between the ratio of GOS net counts to plastic counts and ratio of air kerma to plastic counts. The calibration curve is shown in Figure 3-6. The reference condition was the 120 kV beam at the center-hole location. The correction factor is equal to the first order coefficient of the calculated best fit using a first degree polynomial with no y-intercept function. A noticeable breakdown in the linearity of the curve was observed for the 80 kVp beam resulting from the energy absorption effects as demonstrated by the free-in-air measurements in Table 3-3. The results show that using this method, the correct plastic scintillator response factor may be estimated to within 9% for CT tube potentials of 100 through 135 kVp and all hole locations. Individual percent differences are tabulated in Table 3-4. Each percent difference was calculated by averaging over the 100, 120, and 135 kVp measurements. Column 2 is the percent differences when using the correction method from Figure 3-4. Column 3 is implementing the same except removing the data from hole locations 1 and 3 and averaging those values separately to get one correction factor. The reason for performing this is because these values fall within the 6% variation in mass energy-absorption coefficients and anything outside of this range is due to quenching. The ratio of plastic counts to GOS counts that fell between 0.034
and 0.037 were separated from the regression evaluation since it was deemed their response was independent of quenching effects. Column 4 represents the percent differences by comparing the dosimeter’s response to the average of the top and center hole locations without any type of correction method.

**Angular Dependence**

Figure 3-7 plots the normalized response ratio of dosimeter for axial and normal-axial irradiations. The data was normalized to the 0° measurement on the axial rotation exposure. It should be noted that evaluation revealed the effect on the correction factor which is dependent on the ratio of the scintillators’ response. It is evident that the irradiation from the backend induces a high magnitude of signal loss as well as a more variation in response from the normal-to-axial irradiation, than the axial. This dependence is consistent with the PSD and as noted is less of an issue when the dosimeter is placed in a scatter medium which creates isotropic irradiation that is less directionally dependent.

**Discussion**

Each scintillator demonstrated count rate nonlinearity effects when compared to the ionization chamber. The behavior of both scintillators can be modeled using the nonparalyzable model expressed in Equation 2-7. The results of this model show that, on average, the GOS yielded a deviation from linearity of 5% at 26.3 mGy/sec, while the plastic scintillator demonstrated a 5% deviation from linearity at 324.3 mGy/sec.

Exposure rates of clinical CT units are typically lower than 60 mGy/sec demonstrating the impact of correcting for count rate losses of the gadolinium element. The reported exposure rate dependence of the RC 0.6 small volume ion chamber was ±2% between 0.0005 to 2033 mGy/sec well outside the measured exposure rates that ranged from 2.9...
to 333 mGy/sec. However, this error in dose rate dependence was accounted for by assigning a fixed error for each ion chamber measurement of 5%. When performing this experiment, a source whose output strength is not energy dependent is more ideal for comparing calculated dead times between different energies. The variation in dead time measurements for the GOS scintillator is minimal, and when including all energies the dead time was calculated to be 48 ns. Assigning a dead time value to the plastic scintillator is too arbitrary given the vast uncertainties in the measurements. It is estimated that the actual value would fall between 80 to 140 ns. However presented with an x-ray tube with a more intense beam current a precise value could be determined. One reason for not using a megavoltage therapy accelerator or orthovoltage machine was the introduction of Cerenkov radiation above 150 keV.

An interesting result is that despite a very long decay time, on the order of microseconds, the GOS element demonstrated a dead time on the order of nanoseconds. According to Knoll, when a pulse of radiation is significantly larger than the dead time of the PMT, which is certainly acceptable to model hundreds of microseconds decay function as a long pulse on the nanosecond second scale, the steady-state models may be applied with reasonable accuracy. Also noteworthy, as the dose rate was increased, the gadolinium counts were still accumulating after the radiation was stopped insinuating that the decay time increased at higher dose rates.

The energy response of the plastic scintillator was most affected at depths beyond 16 cm showing a decrease in sensitivity. This was attributed to ionization quenching effects as the electron spectrum is significantly lower as more of the primary beam is attenuated and scatter becomes predominant. The GOS scintillator remained
accurate relative to the ionization chamber among all depths within 13%. The effect of quenching however was less apparent in the free-in-air measurements citing a small variation in the calculated quenching factor. It appears that GOS, when properly corrected for dead time, is a feasible dosimeter in the higher radiological energy range along with other high Z scintillators, and the use of solely a plastic scintillator may require more correction factors.

A dual scintillator dosimeter serves an acceptable tool for CT dosimetry demonstrating a superior response to only using a plastic or GOS sensitive element. Using the response ratio shows no significant variation from beam direction and high reproducibility. Each scintillating element individually is flawed to some degree, as documented in this manuscript, limiting the minimum energy dependence that can be achieved. However, when using both scintillators these flaws can be overcome and the energy dependence can be minimized. Nine percent energy dependence between 100 and 135 kVp is certainly promising and correlating the ratio of the response may even more accurately correct air kerma measurements into dose to relevant bodily tissues.

The biggest advantage of using a dual scintillator dosimeter is that the quenching effect from polystyrene may be minimized and can be potentially used to correct the response of other PSDs used in an array for rapid dosimetry.

With the capability of acquiring data in real-time, an analog control program can be coded to measure depth-in-phantom correction factors for different angular projections of the CT x-ray tube as it rotates around the table. Since the dosimeter is fixed, different gantry angles will result in varying depths that the x-ray beam must penetrate before reaching the dosimeter. The full rotation time of a CT x-ray tube is
somewhere between 0.3 to 0.5 seconds. With the detection system utilized in this investigation, counting data can be obtained within 10 ms intervals. If the acquisition time interval on the PMT was set to 50 ms, 6 to 10 depth dose correction factors could be generated in a single axial tube rotation. For helical scans, the number of angular projections correlates directly with the pitch and the PMT gate time. This type of data acquisition technique would be very useful for tube-current modulation dosimetry studies.

**Conclusion**

This study addressed the count rate effects of using high-Z fiber-optic dosimeters for dosimetry purposes, which previously were interpreted as having a strong energy dependence that was attributed to x-ray absorption properties. By using a high-Z scintillator along with a plastic scintillator, ionization quenching effects may be limited. The dosimeter investigated in this study provides a correction method to offset the depth and energy response of a plastic scintillator used in medical x-ray imaging dosimetry studies. Incorporating this type of photon detection system with fiber-optics allows real-time corrections to be made as the gantry rotates around the phantom exposing the dosimeter to the different irradiation conditions characterized in this study. In addition, it proves to be very useful in projection imaging dosimetry where the beam may be highly attenuated when it reaches the dosimeter. Although it doesn’t perform as well as an ionization chamber, the dosimeter’s small size makes it an ideal candidate for physical phantom dosimetry and situations requiring a high spatial measurement without a loss in sensitivity. This would be useful for beam profile measurements or low-level background measurements.
After a review of commercially available scintillators it is recommended that calcium fluoride is the ideal high Z scintillator due to its smaller effective atomic number (≈16.5), which is very similar to bone (≈13), and minimal quenching characteristics (kB=0.0052). However, it is a difficult and expensive process to integrate with an optical fiber while preserving the small diameter. It would be most advantageous to be fabricated into a fiber with a surrounding cladding material similar to the plastic fibers. Scintillators with low-energy k-edges are ideal because the k-edge will induce a discontinuous relationship when taking the ratio of both scintillators which would make developing a correlation very difficult.

The effects of quenching make the dual scintillator detector less ideal for evaluating the beam energy. Although this relationship is relatively linear it still is less than optimal. In the conclusions section of this dissertation, two candidates are brought to attention as possible scintillators that would be most ideal for a dual scintillation detector. The theory of using two scintillators to model the response of biological tissues is very much proven given the proper selection of both scintillators. In addition, the use of plastic scintillators as being more tissue-equivalent may be overstated given the gadolinium oxysulfide scintillator out-performing the plastic in terms of energy and depth dependence.
Table 3-1. Common properties of GOS and polystyrene

<table>
<thead>
<tr>
<th>Property</th>
<th>Gadolinium oxysulfide</th>
<th>Plastic scintillator (BCF-12)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Molecular Formula:</td>
<td>Gd₂O₂S</td>
<td>C₈H₈</td>
</tr>
<tr>
<td>Peak Emission Wavelength (color):</td>
<td>545 nm (green)</td>
<td>435 nm (blue)</td>
</tr>
<tr>
<td>Decay Time:</td>
<td>&gt; 200 µs</td>
<td>3.2 ns</td>
</tr>
<tr>
<td>Rise Time:</td>
<td>Few ns</td>
<td>≈ 1 ns</td>
</tr>
<tr>
<td>Effective Atomic #:</td>
<td>59.5</td>
<td>5.74</td>
</tr>
<tr>
<td>K-edge:</td>
<td>50.22 keV</td>
<td>&lt; 1 keV</td>
</tr>
<tr>
<td>Mean Excitation Energy:</td>
<td>493.3 eV</td>
<td>68.7 eV</td>
</tr>
<tr>
<td>Absorption Photon Yield:</td>
<td>60,000 γ/MeV</td>
<td>8,000 γ/MeV</td>
</tr>
<tr>
<td>Specific Gravity</td>
<td>7.32 g/cc</td>
<td>1.05 g/cc</td>
</tr>
</tbody>
</table>

Table 3-2. Polystyrene dead time and correction factor evaluation from free-in-air measurements and Monte Carlo methods

<table>
<thead>
<tr>
<th>X-ray spectrum</th>
<th>Scintillator dead time</th>
<th>Air kerma sensitivity</th>
<th>Mass-energy absorption</th>
<th>Scintillator dose sensitivity</th>
<th>Quenching factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>Effective energy (keV)</td>
<td>τ (ns)</td>
<td>⅛ε (nGy per count)</td>
<td>μₑn Poly / ρ Air</td>
<td>S_sys/m (10⁶ counts per Gy)</td>
<td>kB = 0.016</td>
</tr>
<tr>
<td>39.6</td>
<td>849.6 ± 469.3</td>
<td>821.0 ± 3.10</td>
<td>0.467 ± 0.00014</td>
<td>2.609 ± 0.067</td>
<td>0.720</td>
</tr>
<tr>
<td>47.5</td>
<td>39.46 ± 177.9</td>
<td>695.9 ± 3.37</td>
<td>0.534 ± 0.00010</td>
<td>2.689 ± 0.069</td>
<td>0.737</td>
</tr>
<tr>
<td>53.9</td>
<td>246.7 ± 118.2</td>
<td>637.8 ± 3.87</td>
<td>0.596 ± 0.00012</td>
<td>2.630 ± 0.073</td>
<td>0.746</td>
</tr>
<tr>
<td>59.1</td>
<td>144.2 ± 75.86</td>
<td>618.4 ± 3.90</td>
<td>0.647 ± 0.00011</td>
<td>2.503 ± 0.069</td>
<td>0.750</td>
</tr>
<tr>
<td>62.3</td>
<td>94.36 ± 54.02</td>
<td>601.3 ± 3.98</td>
<td>0.671 ± 0.00015</td>
<td>2.487 ± 0.071</td>
<td>0.756</td>
</tr>
</tbody>
</table>
Table 3-3. GOS dead time and correction factor evaluation from free-in-air measurements and Monte Carlo methods

<table>
<thead>
<tr>
<th>X-ray Spectrum Effective energy (keV)</th>
<th>Scintillator Dead Time τ (ns)</th>
<th>Air kerma sensitivity $\gamma \varepsilon$ (nGy per count)</th>
<th>Mass-energy absorption $\mu_{en,Pol}^{Poly} / \rho_{Air}$</th>
<th>Scintillator dose sensitivity $S_{sys/m} (10^6 \text{counts per Gy})$</th>
<th>Quenching factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>39.6</td>
<td>60.75 ± 2.414</td>
<td>2.203 ± 0.044</td>
<td>64.43 ± 0.036</td>
<td>7.045 ± 0.125</td>
<td>0.993</td>
</tr>
<tr>
<td>47.5</td>
<td>54.10 ± 1.244</td>
<td>1.849 ± 0.040</td>
<td>77.44 ± 0.012</td>
<td>6.983 ± 0.149</td>
<td>0.993</td>
</tr>
<tr>
<td>53.9</td>
<td>50.62 ± 0.587</td>
<td>1.726 ± 0.034</td>
<td>83.13 ± 0.194</td>
<td>6.97 ± 0.140</td>
<td>0.993</td>
</tr>
<tr>
<td>59.1</td>
<td>49.65 ± 0.354</td>
<td>0.5890 ± 0.033</td>
<td>82.97 ± 0.324</td>
<td>7.099 ± 0.134</td>
<td>0.994</td>
</tr>
<tr>
<td>62.3</td>
<td>48.58 ± 0.364</td>
<td>1.786 ± 0.0031</td>
<td>79.25 ± 0.387</td>
<td>7.066 ± 0.126</td>
<td>0.994</td>
</tr>
</tbody>
</table>

Table 3-4. Response factor comparison between estimated air kerma conversion coefficient vs. measured. Each position averages the percent deviation from the measured for 100, 120, and 135 kV energies.

<table>
<thead>
<tr>
<th>Depth into phantom (cm)</th>
<th>Corrected by using all data points</th>
<th>Corrected by using selective data points</th>
<th>No Correction (response factor determined by middle phantom location)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.66</td>
<td>5.29% ± 1.31%</td>
<td>0.38% ± 1.27%</td>
<td>0.38% ± 1.27%</td>
</tr>
<tr>
<td>9.67</td>
<td>4.90% ± 2.10%</td>
<td>1.36% ± 2.03%</td>
<td>5.83% ± 1.29%</td>
</tr>
<tr>
<td>16.0</td>
<td>2.23% ± 0.72%</td>
<td>0.36% ± 0.65%</td>
<td>0.36% ± 0.65%</td>
</tr>
<tr>
<td>22.35</td>
<td>2.62% ± 0.62%</td>
<td>0.84% ± 0.62%</td>
<td>22.2% ± 3.30%</td>
</tr>
<tr>
<td>30.35</td>
<td>7.85% ± 1.31%</td>
<td>11.4% ± 1.33%</td>
<td>247.9% ± 104.4%</td>
</tr>
</tbody>
</table>

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Figure 3-1. Size of the dual scintillator dosimeter relative to a United States quarter. Photo courtesy of Matthew R. Hoerner.
Figure 3-2. Count rate effects from 120-keV x-ray exposure to the GOS scintillator (a) and the plastic scintillator (b). The line represents a fit of the data to a second degree polynomial to demonstrate the nonlinearity of the GOS. The calculation of dead time for both the GOS scintillator (c) and the plastic scintillator (d) are illustrated using the first degree polynomial best fit.
Figure 3-3. Electron fluence spectrum through the polystyrene scintillator for various hole locations in the CTDI phantom from exposure to a 120-kVp CT x-ray beam. Data acquired using Monte Carlo methods. CTDI nested head and body phantom are shown depicting the position of each hole location relative to the x-ray tube fixed at the 12 o’clock position in the diagram.
Figure 3-4. Gadolinium oxysulfide (green line) and polystyrene (blue line) relative photon fluence spectrum from a 120-kVp CT x-ray beam. Hole locations 1 and 3 correspond to 1 (solid lines) and 16 (dashed line) cm respectively. Data acquired using Monte Carlo methods. Note the polystyrene fluence spectrum y-axis on the left and gadolinium oxysulfide spectrum y-axis on the right.
Figure 3-5. Gadolinium oxysulfide and plastic scintillator energy response as a function of depth into CTDI phantom for three tube potentials: 100, 120, and 135 kV. Three sets of data are plotted to include gadolinium oxysulfide corrected using a dead time of 48 ns, uncorrected gadolinium oxysulfide, and the plastic scintillator. The dashed lines correspond to the +/- 10% deviation from the normalization point. The data were all normalized to the 120 kV center-hole data points, which is positioned at 16 cm depth.
Figure 3-6. Determination of correction factor for the plastic scintillator for energies 80 to 135 kV. Correction factor was calculated by taking the response factor of each scintillator and normalizing to the calibration condition which was defined at 120 kV beam at the center-hole location. Solid line represents the linear no threshold fit and the two dashed lines represent the standard deviation of the fit. Top Figure (a) shows the entire range of data. Bottom Figure (b) is the range of data (within the grey box) that were deemed unaffected by quenching.
Figure 3-7. Angular dependence of the response ratio of the dosimeter to an axial irradiation (a) and normal-axial irradiation (b). In both cases, the data has been normalized to the zero-degree axial irradiation response. The schematic to the right of each plot illustrates the experimental setup with the shaded circle representing a head on view and the square a side view of the scintillator. The white rectangle indicates the optical fiber, which is not visible in the top diagram.
CHAPTER 4
CHARACTERIZATION OF A LINEAR FIBER-OPTIC DOSIMETER FOR COMPUTED TOMOGRAPHY

Introduction

The convention for computed tomography (CT) physical phantom dosimetry is performed with dosimeters that have a small sensitive volume to satisfy cavity theory specifications and be integrated easily within phantoms. In radiological dosimetry, meeting cavity theory requirements is easily achievable due to a small wall thickness required to provide sufficient build-up to create full charge particle equilibrium and localized charged particle energy deposition. The principle issue in radiological dosimetry is matching the material composition of the dosimeter to replicate the photon energy-absorption characteristics of the biological tissue of interest. Ionization chambers are the gold standard in radiological dosimetry, but are not feasible for internal organ dosimetry due to their large size which limits their spatial resolution and makes them difficult to integrate within phantoms. This has prompted the development of dosimeters that are small enough to be easily integrated within all types of phantoms.

Dosimetry using physical phantoms has been performed using film, or more recently, solid state dosimeters such as TLDs and OSLs. Film is very useful for measuring dose to an area, but is energy dependent and non-linear. OSLs and TLDs require a post-irradiation process to measure the energy stored within each crystal. Commercially available OSLs and TLDs are made with a small sensitive volume suitable for point dosimetry. For example, the size of an OSL nanoDot™ (LANDAUER, Glenwood, IL) is 10x10x20 mm³.
When investigating dosimeters that cannot perform absolute measurements, it may be more advantageous to research a dosimeter with a sensitive volume that represents the geometry of the region of interest to obtain spatially-integrated measurements. For example, a thin layer of film can be used to simulate the absorbed dose of a layer of epithelial cells. In CT dosimetry, dosimeters are placed within phantoms that have segmented internal organs. Organs are quite voluminous relative to the size of the sensitive volume of point dosimeters being utilized in radiological dosimetry studies. Therefore multiple readings are made and averaged to obtain an organ dose measurement. The method of delivering radiation in CT which scans in the table translation direction, allows for partial irradiation of critical anatomy. In addition, with the use of tube-current modulation, the intensity of the x-ray beam is highly variable along direction table translation. These considerations demonstrate the value of a dosimeter capable of measuring cumulative z-axis dose in CT dosimetry studies.\textsuperscript{119} Such a detector does exist though, the 10-cm pencil ionization chamber. The detector evaluates the axial dose within a cylindrical CTDI phantom, exactly as the methodology discussed in Chapter 1. However, the detector is unsuitable for in-phantom dosimetry due to its large size, as well as no formulism for evaluating dose since cavity theory would not be applicable to this detector.\textsuperscript{84,120}

**Fiber-Optic Dosimeters**

Fiber-optics may permit the development of such a dosimeter. Plastic scintillators that have been investigated for photon dosimetry are fabricated into miniature cylindrical fibers with a surrounding cladding material. This allows light to travel within the fiber through internal reflection, and function jointly as a detector and optical waveguide. When the scintillating fiber is placed orthogonal to the direction of
incident radiation, the small diameter of the fiber maintains charged particle equilibrium. The ideal dosimeter would demonstrate uniform spatial sensitivity across its active length. There are four main issues when considering the spatial sensitivity of a scintillating fiber that may lead to nonuniform spatial sensitivity effects.

1. Geometric coupling efficiency
2. Reflective vs. non-reflective efficiency of the reflector
3. Reflection efficiency at the uncoupled end of the scintillating fiber
4. Attenuation of light at various positions in the scintillating fiber

Number 1 considers the loss in spatial sensitivity from misalignment between the scintillator and the optical fiber inducing a geometric spatial efficiency effect. This effect arises if the diameter of the scintillator is larger than the diameter of the optical fiber, leading to escaped light photons. In addition, any misalignment between the two fibers during construction could potentially magnify this effect. Number 2 explains the effect of differences in sensitivity of the fiber at the surface of the optical fiber where light can be transmitted without being reflected within the scintillator compared to light that must be reflected. In this case, the transmission cone, defined by the critical angle of internal reflection within the scintillator, is greater than the acceptance cone, defined by the numerical aperture of the scintillator-waveguide surface. Number 3 accounts for the uncoupled end of the fiber reflecting less than the entire incident light. If the reflector was 100% efficient, then transmission losses would be negligible because light would travel the same distance when assuming isotropic light emission. Therefore, any reflector with less than 100% light reflection efficiency will be less sensitive to detecting light emitted on the reflector side of the fiber. Number 4 includes the length of fiber and the attenuation coefficient, and evaluates their effect on the spatial sensitivity.
Plastic Scintillating Dosimeters

Plastic scintillating dosimeters (PSDs) have been used extensively for radiological dosimetry applications. Hyer examined the utility of PSDs in the CT energy range and estimated that measurement uncertainties for evaluating organ doses were no more than 4%. Contributing factors included energy dependence due to changes in beam quality and scatter, angular dependence from a full normal-to-axial irradiation in a scatter medium, reproducibility, and variation in response from bend radius. Most importantly, these dosimeters operate in real time and because of their small size and flexibility can be easily used with phantoms providing the user with accurate and rapid evaluation of organ doses. The fabrication of these dosimeters for use in CT can be achieved by simply modifying the length of the scintillating to any desired length that would cover the organ or target volume desired.

Materials and Methods

Simulated Spatial Sensitivity

Simulations were conducted to investigate the applicability of an elongated plastic scintillating fiber by evaluating any variation in spatial sensitivity. The plastic scintillating fiber and optical waveguide used in this study were commercially available products with physical and optical properties listed in Table 3-1. A simple calculation reveals that the transmission cone of the scintillating fiber (42.74°) is greater than the acceptance cone of the optical fiber (15.81°). This satisfies the third condition since unreflective light reaching the scintillator-waveguide boundary will be transmitted to the optical fiber with the same efficiency as reflected light. Light propagation in fiber-optics is often restricted to analysis of meridional rays, despite both skew and meridional rays traveling within a fiber-optic cable. This type of analysis was justified by Cozannet and
Treheux\textsuperscript{122} provided the appropriate acceptance angle of skew rays is used. Meridional rays are less complex since the ray is confined to the plane containing the fiber’s optical axis and the point at which the ray originated. Skew rays travel in a non-planer zig-zag path that never crosses the axis of the fiber. Knowing the numerical aperture of the optical fiber, the core refractive index of the scintillating fiber, and optical fiber; the effective numerical aperture and effective acceptance cone of the optical fiber and scintillator is defined in Equation 4-1.

\[
NA_{eff} = \sin \theta_a = \frac{n_{core}^{opt}}{n_{core}^{scint}} \cdot NA_{opt} \tag{4-1}
\]

Equation 4-1 was referenced from Beddar et al.\textsuperscript{107}, which was used to model the light collected from the acceptance cone of the scintillator-optical fiber coupling. A MATLAB (Mathworks Inc., Natick, MA) script function was written to evaluate the spatial efficiency of the scintillator for different lengths and attenuation coefficients. Equation 4-2 formulates the fraction of light attenuated by the scintillator at different distances from the coupling surface.

\[
F(x) = \frac{1}{2} \left[ \int_{0}^{\theta_a} e^{-\lambda x \sec \theta} \, d\theta \right] + \frac{1}{2} \varepsilon \left[ \int_{0}^{\theta_a} e^{-\lambda (2L-x) \sec \theta} \, d\theta \right] \tag{4-2}
\]

The term L specifies the length of the scintillating fiber, and \( \lambda \) describes the linear photon attenuation coefficient. The integral is independent of radial distance from the central fiber axis because, although it will induce maximum one additional reflection, it does not change the total distance traveled within the scintillator. The exponential is integrated over all angles less than the acceptance angle defined in Equation 4-1.

A separate script, modeling the light loss from fiber alignment and reflector efficiency, was coded to determine any axial spatial dependence from a uniform
exposure. The method of evaluating the alignment effects involved calculating the success that a light ray emitted at a position within the phantom at a specific angle would cross the fiber boundary directed into the core of optical waveguide. This type of analysis is illustrated in Figure 4-1. The coupling efficiency at a position \((x)\) may be evaluated by integrating over the passable locations and the non-passable locations designated by the white and dark areas respectively. The location the light ray crosses the surface is determined by its \(y\) position. The \(y\) position can be determined by the angle it is emitted and the distance from the surface as seen in the schematic. Light contributions from the cladding material were ignored due to complexity. At each position \((x)\) the probability \((p)\) may be evaluated by integrating over \(p(y)\), which is expressed in discrete voxels which represent discrete radial positions \((y)\). The chosen intervals were 0.005 mm. For simplification, a voxel along the axis was assigned a number 1 if it was within the white region or 0 if it was in the dark region.

**Construction**

The linear fiber-optic dosimeter (LFOD) was constructed by modifying the sensitive element of the fiber-optic coupled (FOC) dosimetry system. The plastic scintillating fiber (BCF-12, Saint-Gobain Crystals, Nemours, France), which was previously 3 mm in length, was replaced to mirror the dimensions of organs for the University of Florida adult anthropomorphic phantom. Therefore, a series of scintillating fibers were fashioned to variable lengths representing the typical dimensions of an adult human organ: 3.0, 5.0, 7.0, 10.0, and 15.0 cm. One end of the scintillating fiber was coupled to a multi-mode transmission optical fiber (400-UV, Polymicro Technologies, Phoenix, AZ), having a 400-μm diameter, and two meters in length. Before coupling, both ends were polished to improve coupling efficiency. The remaining end of the
scintillator was coated with reflective paint (EJ-510, Eljen Technology, Sweetwater, TX) to increase the acceptance efficiency of the fiber-optic system. The coupled scintillating-transmission fiber assembly is encased in heat-shrink tubing, mounted to a SMA connector (SMA-490, Ocean Optics Inc., Dunedin, FL) and sealed to prevent ambient light contamination. The area of heat-shrink tubing covering the scintillating fiber was spray painted to identify the sensitive element.

A separate LFOD was constructed to include a reference fiber for dosimetry measurements. This was necessary to quantify the stem counts generated in the optical fiber from the natural florescence when exposure to ionizing radiation. In order to calibrate and quantify the air kerma accurately the dosimeter was made with a 10-cm scintillating fiber, the same active length as a 10-cm pencil chamber ionization chamber.

Characterization

General characterization

The characterization of the LFOD for spatial sensitivity, energy dependence, reproducibility, and linearity was performed using a clinical radiographic x-ray tube (CPI Indico Plus, Communications & Power Industries International Inc., Georgetown, Ontario, Canada) with variable output and characterized half-value layer of 3.66 mm Al at 80 kVp. All measurements were repeated five times to reduce statistical uncertainties and provide reliable reproducibility measurements. A 100-mm pencil ionization chamber and associated electrometer (chamber model RC3CT, Accu-Dose electrometer model, RadCal Corporation, Monrovia, CA) were used to provide simultaneous exposure measurements.

The linearity and reproducibility of the two LFODs were evaluated simultaneously by delivering a variable exposure by increasing the tube-current product from 32 to 400
mAs. The modification of the tube-current product was done by varying both the exposure time and tube-current. In effect this method also represented the dose rate linearity of the dosimeter. However, from the previous chapter the dose rate of the dosimeter was been well characterized. A reference chamber (pencil chamber, with cap) was used to quantify the reference air kerma for each tube setting. The detectors were placed on top of a backscattering medium (PMMA) at 40 cm from the anode. The field size was collimated to 5cm x 5cm at the detector measurement location to ensure the x-ray beam was contained by the sensitive element of the dosimeter and ion chamber.

Energy dependence was evaluated by delivering x-rays from a range of tube potentials at the LFOD and reference chamber positioned on the central axis free-in-air. Several x-ray tube voltages of 80, 90, 100, 110, 120, and 130 kV were selected for this study. The pencil chamber, with the cap, and LFOD were centered along the cross hairs of the x-ray tube and placed orthogonal to the axis that runs parallel to the variable x-ray intensity to avoid the heel effect. The x-ray field was collimated to a 7cm by 7cm field at the detection depth (40cm SDD). The energy dependence was quantified by taking a ratio of the LFOD readout to the measured air kerma, expressed in counts/mGy, and plotting this value normalized to the 80 kV correction factor for each tube potential.

For completeness in evaluating the dosimeter’s response to changes to beam hardening, varying thicknesses of soft-tissue equivalent material were placed on top of both the dosimeter and pencil chamber, with the cap, to evaluate effects from beam hardening and scatter. Two custom rectangular tissue equivalent phantoms were
constructed to house the ion chamber and dosimeter and a set of rectangular slabs. Figure 4-2 illustrates the detector phantom with slabs of soft-tissue equivalent material slabs placed on top. Exposures were delivered at 120 kV, 50 mAs, and a fixed field size of 5x5 cm. which was measured at the detector depth (60cm SDD) to ensure the full beam, comprised of scatter and primary x-rays, would be contained by the sensitive volume of each detector. Similar to the previous experiment, a plot was created containing the normalized detector sensitivity in counts/mGy, versus depth in the material.

Spatial sensitivity was evaluated by sampling counts as a function of length exposed to a uniform x-ray field. A step motor (Model 1000, Advanced Radiation Measurements, Port Saint Lucie, FL) was used to insert each LFOD incrementally into the x-ray field generated by a clinical portable x-ray unit at 80 kVp and 10.5 mAs. The LFOD’s were placed in the phantom with 1.5 cm of material to attenuation and backscatter. Sensitivity was measured in number of measured counts as a function of exposed scintillating fiber length (sensitive region) in 5 mm increments. Variable spatial intensity from the heel effect was avoided by inserting the LFOD into the x-ray field perpendicular to the axis containing the heel effect.

Characterization for computed tomography dosimetry

The effects of spatial sensitivity in a phantom from a CT irradiation were evaluated. The 100-mm pencil chamber was used to quantity reference air kerma values from varying beam profiles delivered along the axis of rotation which was aligned with the center-hole location of a CTDI body phantom. Beam profiles were varied by adjusting the collimation width. Exposures were delivered from a Toshiba Aquilion One volumetric scanner (Toshiba America Medical Systems Inc., Tustin, CA) in a single
rotation with no table translation to each detector free-in-air with variable collimation width of the x-ray beam along the axis of rotation while keeping the exposure parameters held constant. The maximum selected collimation width was 8 cm to include the fall-off from scatter and attenuation through the bow-tie filter. The measurement was performed free-in-air because of the wide profiles associated with in-phantom measurements attributed to scatter radiation.\textsuperscript{8} Boone demonstrated through computational methods that efficiency of measuring CTDI\textsubscript{vol} is quite low after 10-mm slice acquisitions. Therefore the free-in-air measurement technique was selected as the optimal method of comparing air kerma values from the 100-mm pencil chamber and the LFOD.

A unique characterization method was required to characterize the LFOD for CT dosimetry to approximate the actual irradiation conditions experienced by the dosimeter. For CT dosimetry characterizations, a CTDI body phantom provided the most realistic physical representation of the body’s attenuation characteristics while also able to accommodate an ionization chamber to quantify reference air kerma for the 100-mm LFOD. The effects of penetration depth, beam energy, scatter radiation, and spatial sensitivity were all evaluated simultaneously by quantifying the ratio of detected light photons to air kerma from different beam energies. For this characterization a different reference chamber was selected. CTDI pencil chambers are acceptable cavity-type chambers as long as the beam (primary and scatter tail) is confined within the sensitive volume of the detector. However, for most clinical scans the scanning region will fall outside of the 100-mm chamber and the correlation between air kerma and dose to soft tissue will breakdown.
To evaluate the dose within the phantom the AAPM protocol for evaluation of CT dose was adopted. From this protocol the axial dose profile is saturated by performing a full scan with table translation using a narrow beam over an elongated CTDI phantom with a detector centered on the machine isocenter and placed at the midline of the phantom. Figure 4-2 illustrates the setup geometry and response of the LFOD. By using a narrow collimation width and a long phantom, the integrated dose will approach an equilibrium value and generate a flat dose profile over the center-hole location within the phantom. This equilibrium dose is characteristic of the tube voltage, mAs, filter shape, and pitch. In this scenario, the tube voltage was varied to produce a energy-dependent response at the detector position that includes a primary and scatter component. The phantom was assembled using a series of adjacent CTDI body phantoms and a pair of tissue-equivalent cylindrical phantoms on each end. Measurements were taken for three tube voltages (100, 120, and 140 kV) at the center hole location. Equilibrium was established for LFOD when the integral counts at later time intervals approached zero at the beginning and end of the scan.

**Results**

**Computational Simulations**

Each MatLab simulation was performed to vary the highlighted parameters that were deemed significant to affect the spatial sensitivity of the LFOD. The default parameters for fiber length, attenuation, reflector efficiency, and alignment were taken from the manufacture information for the s components highlighted in Table 4-1: 10 cm, 0.00370 cm\(^{-1}\), 0.95%, and completely aligned. Figure 4-4 shows the effects of the reflector efficiency and effects from fiber alignment. Figure 4-5 shows the effects of fiber length and attenuation. These simulations indicate a higher importance of fiber
alignment, and a lesser importance of reflector efficiency. The fiber light attenuation coefficient plays an important role as well, especially at greater lengths. The specified attenuation coefficient by the vendor was 270 cm\(^{-1}\) indicating that fiber alignment plays the most significant role in determining the spatial independence of these types of detectors. In general, this effect is most observable within the first 1-2 cm of the fiber from the coupling side of the scintillating fiber.

**Linearity and Reproducibility**

Table 4-2 summarizes the average readings and standard deviation sampled from 5 measurements, over the exposure range delivered from x-ray tube current-time products from 32 to 400 mAs. The data was fit to a linear no-threshold function to determine the sensitivity of each fiber. At the bottom of Table 4-2 the linear correlation coefficient is displayed to evaluate the linearity of the data along with the sensitivity expressed in terms of counts per air kerma (mGy).

**Energy Dependence**

The 10-cm LFOD containing the reference fiber to quantify background radiation was used to match the length of sensitive element to the pencil ion chamber. The normalized readings, in counts/mGy, were plotted against the selected tube potential. The data is displayed in Figure 4-6. The response was normalized to the 120 kV data point. The energy response from changes to tube voltage is shown to be positively linear, with an average 0.6% increase per kV. For reference, the PSD using a 3mm scintillator was characterized as a positive energy dependence of 0.68% per kV offering a similar free-in-air energy response.

The depth/penetration dependence of the LFOD is shown in Figure 4-7. Each reading was normalized to the surface (zero penetration depth) data point to illustrate
the positive response with increasing tissue penetration. This positive increase varied 11% over the 14 cm of soft-tissue equivalent material. The positive slope has been attributed to beam hardening which results in an x-ray beam of higher average energy. At energies between 40 and 70 keV, the ratio of mass energy-absorption coefficients between polystyrene and air is positively linear as well. The reported variation in the PSD was described as 10% across the 16 cm of material. Overall, the energy response of both dosimeters is very comparable indicating little importance on scintillator length and inaccuracies associated with beam energy.

**Spatial Sensitivity**

Spatial sensitivity was evaluated for the series of LFODs by evaluating the response to the length of scintillating fiber irradiated in the x-ray field. The six scintillators, with sensitive element lengths of 3.0, 5.0, 7.0, 10.0, 12.0, and 15.0 centimeters were all evaluated. A step motor was used to insert incrementally into the x-ray field axis perpendicular to the heel effect. The x-ray field was generated by a clinical portable x-ray unit at 80 kVp and 10.5 mAs. An ion chamber verified the uniformity of the area of the field used for exposures to within 5%. Sensitivity was measured in number of measured counts as a function of exposed scintillating fiber length (sensitive region) at 0.5 cm intervals. A linear fit was applied in order to obtain the number of counts per centimeter of exposed fiber and along with the linear correlation coefficient. The linearity of the LFODs proved to be excellent demonstrating good uniform spatial sensitivity. The lowest linear correlation coefficient was calculated to be 0.9933. Each detector’s response as a function of exposed length is plotted in Figure 4-7.
Characterization for Computed Tomography Dosimetry

Figure 4-9 shows the response of the ion chamber and LFOD to varying beam collimation widths irradiated free-in-air. The tube voltage was set to 120 kV, the tube-current to 200 mA, and the slice thickness was maintained to 0.5 mm. The beam width was changed by selecting variations of number of slices per rotation. For example, a 64 slice scanner, with a 0.5 mm slice thickness would use a 3.2-cm beam. The beam width is measured along the isocenter, which corresponds to the center-hole location. Beam collimation thicknesses ranged from 0.5 mm to 7 mm in order to contain the entire beam within the sensitive length of each detector. The response by the LFOD was exceptional, again proving no spatial sensitivity effects.

Scan parameters were as follows. The full scan was performed over 43 cm of acrylic and soft tissue-equivalent material at a pitch of 1.5. The beam was collimated to 20 mm in order to perform the helical scan with a tube current of 200 mA, 0.8 s rotation time, and 30 mm/rot table translation. The tube voltage was selected from 100, 120, or 140 kV using the medium body filter. Data was captured using the detection system to record signal at each time interval (0.1s) in addition to the total number of counts detected. Three measurements were taken at each tube voltage and then combined together and averaged to generate three profiles. The time scales were aligned together to create a time index in order to account for the scans starting at different times. Each scan was 12.9 seconds, therefore 129 measurements contributed to the profile. Figure 4-10 plots the profile from each tube voltage as well as a fit function for all three profile using a Gaussian distribution. The profile is actual a relative profile taken by normalizing the counts in each bin by the maximum for that energy. The Gaussian distribution is actually a two component Gaussian, one modeling the scatter
radiation contribution, and the other for primary beam radiation. The two component Gaussian distribution which includes the primary and scatter component is plotted in Figure 4-11A. Irradiating over the entire scan length results in 50% of accumulated/equilibrium dose is from scatter, but at the center, when the beam is directly irradiating the detector, only 30% is of the dose is from scatter. These implications may affect how detectors are calibrated moving forward in CT.

To demonstrate the effect of scatter, Table 4-3 shows the relative air kerma correction factors (normalized to 140 kV) at the center hole location of the CTDI phantom for the gadolinium scintillating detector (GSD) described in Chapter 3 and the correction factors for the LFOD. The correction factor for the LFOD varies by approximately 0.3% per kV which is half of the free-in-air energy dependence of 0.6% per kV. The gadolinium detector performs extremely well, especially when compared to the LFOD, which can be attributed to the majority of scatter radiation that it receives compared to the primary radiation. This is also illustrated in Figure 4-11B which shows the broadness and magnitude of the gadolinium scatter profile compared to the primary curve. Scatter radiation, or low energy radiation, has very similar energy absorption characteristics among materials because x-ray interactions are dominated by photoelectric absorption which results in all of the x-ray energy absorbed locally.

The energy dependence of the LFOD was correctable by using the scatter and primary profile as well as the measured air kerma to generate a primary and scatter correction factor. The count data indexed at each time step was binned into scatter radiation and primary radiation based on the Gaussian distribution in Figure 4-11 A. The counts in each bin were summed to get net scatter counts and net primary counts.
Using the measured air kerma from the ionization chamber, a MatLab program was constructed to find the optimal values of the scatter and primary correction factors. The primary correction factor was energy dependent and the scatter coefficient was the same for each producing four correction factors. These correction factors are listed in Table 4-3 in the far right column. Using these correction factors and the LFOD data, the air kerma could be estimated to less than 1%.

**Discussion**

The LFOD demonstrates very similar characteristics to the PSD. This includes an exceptional linear response with beam intensity and a positive linear energy dependence within the exposure range of radiographic and CT imaging modalities. The overall energy dependence indicates an increasing linear trend with respect to x-ray tube with an estimated variation in sensitivity of 0.6% per kV. Good spatial sensitivity was observed for each of the LFODs used. The spatial sensitivity did appear to regress with increased length, which was an expected result from the simulations. The geometric coupling effect was observed for a few of the dosimeters, and is indicated at the most distal lengths of the spatial sensitivity evaluation since the scintillator optical-fiber interface were inserted last in the experiment. From the free-in-air characterization, the dosimeters showed no loss in sensitivity with respect to the ion chamber.

One of the benefits of CT is that the dynamic motion of the gantry tends to reduce the effects of penetration depth leading to a decreased variation in sensitivity. This was demonstrated by Fisher who used a 32-cm CTDI body phantom to calibrate an FOC for CT organ dose measurements.\(^{123}\) This was further validated by showing that scatter radiation minimizes the energy dependence of a detector. However the LFOD is
less prone to scatter than a point detector because its sensitive element spans centimeters in length. Therefore, it shows higher energy dependence than a point detector. However, it was demonstrated that along the central axis, that the count profile can be used to correct the dosimeter for energy dependence from the primary x-ray beam by fitting the profile to a dual-component Gaussian distribution and calculating the appropriate correction factors.

The main advantage of this type of detector is that its sensitive length can be used for organ dosimetry by using a scintillating fiber similar to the length of the target organ. This detector will be especially useful in areas of scan termination and initiation as well situations where the surrounding tissue is varying creating different intensity’s along the imaging axis. Most commercial phantoms allow for dose locations in the x-y plane and are separated in the z-plane in some cases up to 10-20 cm. Anatomy changes as well as tube current modulation will lead to dynamic beam profile that cannot be captured using point dosimeters and would be more accurately evaluated using integrated detectors.

**Conclusion**

It was demonstrated that the LFOD exhibits the desirable characteristics of a dosimeter that would be very useful for CT dosimetry. Similar to the PSD, its main drawback is the energy/depth dependence from changes to the x-ray energy spectrum. The LFOD from a general characterization standpoint mirrors the PDS’s response. Both shared an exceptional linear response to beam intensity and a positive free-in-air energy dependence. The LFOD detector geometry allows for a lower variation in detected counts, due to the higher light output from the scintillating fiber. The energy dependence is minimized by using the dosimeter in a high scatter environment versus
free-in-air. Overall, this type of detector appears most appropriate for measuring organ doses, where the scintillating element can be fashioned to the dimensions of the organ being evaluated. The rotating x-ray tube will minimize the angular dependence of the fiber as well as flatten the depth dose profile. In addition to providing rapid and accurate measurements, when properly calibrated, this type of dosimeter demonstrates exceptional utility for in-phantom CT organ dose measurements.
Table 4-1. Optical properties of the plastic scintillator and optical waveguide

<table>
<thead>
<tr>
<th></th>
<th>Plastic scintillating fiber (BCF-12)</th>
<th>Silicon optical fiber (FVP400440480)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cladding</td>
<td></td>
<td></td>
</tr>
<tr>
<td>material:</td>
<td>Acrylic</td>
<td>Silica</td>
</tr>
<tr>
<td>refractive index:</td>
<td>1.42</td>
<td>1.46</td>
</tr>
<tr>
<td>thickness:</td>
<td>15 μm</td>
<td>20 μm</td>
</tr>
<tr>
<td>Core</td>
<td></td>
<td></td>
</tr>
<tr>
<td>material:</td>
<td>Polystyrene</td>
<td>Silica</td>
</tr>
<tr>
<td>refractive index:</td>
<td>1.6</td>
<td>1.44</td>
</tr>
<tr>
<td>diameter:</td>
<td>500 μm</td>
<td>400 μm</td>
</tr>
<tr>
<td>Numerical aperture:</td>
<td>0.58</td>
<td>0.22</td>
</tr>
<tr>
<td>Attenuation coefficient:</td>
<td>2.70 m</td>
<td>8.33 m</td>
</tr>
</tbody>
</table>
### Table 4-2. Reproducibility and linearity measurements with 7-cm and 15-cm linear dosimeters

<table>
<thead>
<tr>
<th>Tissue kerma (mGy)</th>
<th>7-cm FOC linear dosimeter</th>
<th>15-cm FOC linear dosimeter</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean counts</td>
<td>Standard deviation</td>
</tr>
<tr>
<td>1.5</td>
<td>15289</td>
<td>165.12</td>
</tr>
<tr>
<td>1.87</td>
<td>18916</td>
<td>236.45</td>
</tr>
<tr>
<td>2.93</td>
<td>29578</td>
<td>300.22</td>
</tr>
<tr>
<td>3.73</td>
<td>37333</td>
<td>176.96</td>
</tr>
<tr>
<td>4.68</td>
<td>46830</td>
<td>220.57</td>
</tr>
<tr>
<td>5.85</td>
<td>58691</td>
<td>117.38</td>
</tr>
<tr>
<td>7.5</td>
<td>75156</td>
<td>95.44</td>
</tr>
<tr>
<td>18.77</td>
<td>189022</td>
<td>489.89</td>
</tr>
</tbody>
</table>

Sensitivity: 10059 (counts/mGy)  
Corr. Coeff: 0.99999

Sensitivity: 68029 (counts/mGy)  
Corr. Coeff: 0.99955

### Table 4-3. Beam profile data from the full body phantom scan at 100, 120, and 140 kV.

<table>
<thead>
<tr>
<th>Tube Voltage (kV)</th>
<th>Relative air kerma correction factor GSD</th>
<th>LFOD</th>
<th>Total scatter fraction* GSD</th>
<th>LFOD</th>
<th>Scatter fraction along phantom central axis* GSD</th>
<th>LFOD</th>
<th>LFOD correction factors (10000 counts/mGy) Primary</th>
<th>Scatter</th>
</tr>
</thead>
<tbody>
<tr>
<td>100</td>
<td>0.990</td>
<td>0.875</td>
<td>0.832</td>
<td>0.507</td>
<td>0.416</td>
<td>0.301</td>
<td>2.72</td>
<td>0.98</td>
</tr>
<tr>
<td>120</td>
<td>1.002</td>
<td>0.942</td>
<td>0.832</td>
<td>0.507</td>
<td>0.416</td>
<td>0.301</td>
<td>3.57</td>
<td>0.98</td>
</tr>
<tr>
<td>140</td>
<td>1.000</td>
<td>1.000</td>
<td>0.832</td>
<td>0.507</td>
<td>0.416</td>
<td>0.301</td>
<td>4.83</td>
<td>0.98</td>
</tr>
</tbody>
</table>

* Taken from dual-component Gaussian fit where the broad curve represented the scatter contribution and narrow primary.
Figure 4-1. Illustration of the coupling efficiency analysis. At a given (x, y) location within the fiber, lines can be drawn from angles within the acceptance cone to the coupling surface which would result in light rays be transmitted or lost due to the uncoupled areas on the scintillator. The variable x is the length along the fiber which gets emitted at an angle theta. This light rays would fall in a voxel (black or white). The voxel would then shift based on the y position.
Figure 4-2. Tissue equivalent phantoms for depth dose measurements. The left (a) is the LFOD phantom while the right (b) is the pencil chamber’s housing. Both phantoms are placed onto of acrylic back scattering medium. Photo courtesy of Matthew R. Hoerner.
Figure 4-3. Phantom assembly and centered LFOD moving together from the right to left over the scanning length. The plot on the right shows the accumulated dose as a function length scanned. As the scan length increases the integrated dose will approach an equilibrium value. The scan was repeated for the ionization chamber and the equilibrium dose was measured.
Figure 4-4. Relative transmission of light through the optical-scintillating fiber boundary at locations along the scintillating fiber’s axis. The simulation was run to vary the attenuation coefficient of light within the detector ($\lambda$) and the length of the fiber. All other parameters were set to the values listed in Table 4-1.
Figure 4-5. Relative transmission of light through the optical-scintillating fiber boundary at locations along the scintillating fiber’s axis. The simulation was run to vary the efficiency of the reflector ($\varepsilon$) and alignment of the optical fiber and scintillating fiber. All other parameters were set to the values listed in Table 4-1.
Figure 4-6. LFOD response factor measured for varying x-ray tube voltage (80 through 130 kV). Measurements taken free-in-air using a 100-mm pencil chamber to quantify reference air kerma. Error bars correspond to ± 1 standard deviation of the mean.

Figure 4-7. LFOD response factor measured for varying penetration depths into a soft tissue-equivalent medium (0 through 14 cm). Detector position was fixed with 5 cm of backscattering material. Error bars correspond to ± 1 standard deviation of the mean.
Figure 4-8. Evaluation of the 7-cm, 12-cm, 10-cm, and 15-cm LFOD dosimeters for spatial sensitivity using a uniform intensity x-ray field. Linear fit was applied to each data set along with a linear correlation coefficient. Error bars correspond to ± 1 standard deviation of the mean.
Figure 4-9. LFOD response to varying collimation widths. LFOD response was benchmarked by measuring the reference air kerma using a 100-mm pencil ion chamber (mGy). Residual plot below shows the relative error of the linear no-threshold fit to the data. Error bars are shown in to show the +/- 1 standard deviation from the mean.
Figure 4-10. Distribution of counts from discrete time intervals sampled over a helical CT irradiation of a 10-cm LFOD in a custom CTDI phantom. Three of the curves correspond to each of the tube voltage settings and the black line is the two component Gaussian fit.
Figure 4-11. Gaussian fits of each counting profile from (a) LFOD and (b) GSD.
CHAPTER 5
CONSTRUCTION OF AN ANTHROPOMORPHIC PREGNANT FEMALE PHANTOM SERIES FOR COMPUTED TOMOGRAPHY DOSIMETRY

Physical Phantoms for Experimental Radiation Dosimetry

Anthropomorphic phantoms used for dosimetry applications have been an accepted method for measuring organ and tissue doses from medical radiation procedures. The advantage of using this tool for patient dosimetry is the anatomy of human subjects is not required. Anthropomorphic phantoms are used in dosimetry studies primarily to: simulate the human body’s absorption and scatter of ionizing radiation, incorporate dosimeters throughout their volume, and correctly model the position, size, and orientation of the organs of interest residing within the phantom. In general, they offer varying degrees of complexity, with the most complex phantoms designed for a specific imaging modality or representation of a specific anthropometric population. Phantoms built specifically for CT dosimetry are typically constructed in cross sections to accommodate placement of thermoluminescent and optically stimulated dosimeters (TLDs and OSLDs) and to avoid adjoining phantom components in the imaging plane to create a uniform and realistic image. The tissue equivalent materials are characterized to have matching density as human tissue. Custom-built anthropomorphic physical phantoms can be a beneficial alternative to the prohibitive cost of the commercially available CT phantoms and address the specific demands required to produce accurate dosimetry. One of the benefits of building anthropomorphic phantoms is that they can be customized so that dosimeters could be placed along the z-axis (cranial-caudal axis). In the previous chapter, it was demonstrated that linear fiber-optic dosimeters (LFODs) are a good dosimeter for
evaluating dose to large volumes that would otherwise require multiple detectors. Their main advantage is that they can be positioned so that their sensitive length aligns with the z-axis spanning the width of the internal anatomy. This type of detector orientation is not possible using commercially available CT phantoms.

**University of Florida Physical and Computational Phantom Series**

A series of anatomical reference physical phantoms have been fabricated for research applications at the University of Florida (UF).\(^{128}\) The physical phantoms at UF are built slice-by-slice using three classifications of tissue-equivalent substitutes representing the three types of bodily tissue most differentiated by their attenuation characteristics: lung tissue-equivalent substitute (LTES), bone tissue-equivalent substitute (BTES), and soft tissue soft tissue-equivalent substitute (STES), which like other phantoms is a composite material representing the remaining anatomy. The anatomy in each cross section is derived from its virtual blueprint, a computation hybrid phantom provided by Dr. Wesley Bolch’s research group at UF.\(^{129-131}\) A computational hybrid phantom is defined as a voxelized phantom that has been stylized to meet reference criteria which includes: height, weight, and organ mass and volume.\(^{132}\) The shape and location of anatomy is not specified for a reference phantom, but hybrid phantoms maintain a level of accurately since a hybrid phantom uses actual human anatomy providing more realism. Building physical phantoms following the slice-by-slice method has shown to be an exhaustive process taking months to accomplish with extra attention required to each slice from sanding of the BTES, trimming of excess STES material, and engraving time. Previous phantoms required anywhere between 10 to 20 minutes of engraver time for creating a single soft tissue mold.
Research Based Pregnant Female Anthropomorphic Phantoms

Currently there exist commercially available phantoms that anatomically model a reference women or reference male/female for abdominal and pelvis studies. A few have used physical phantoms to perform dosimetry on unborn children in the first trimester. In one study, the abdomen/pelvis region was enlarged to portray the extra weight added over the course of pregnancy. The phantoms utilized by these studies are listed in Table 5-1 with their associated cost. Using an adult reference female phantom will limit measurements to obtaining organ doses, not fetal doses. In the beginning stages of pregnancy this is sufficient. However as the pregnancy progresses and the fetus begins to fully develop it becomes uncontained by the volume of a non-pregnant female uterus, and along with the expansion of mother’s waist and pelvis, the modeling of anatomy of a reference female becomes insufficient. Studies have shown that fetal dose strongly correlates with fetal depth and diameter of the pelvis making pregnant female physical phantoms modeling the later stages in pregnancy more valuable. Other studies have also demonstrated the effect of patient size on organ doses.

Building a Novel Anthropomorphic Phantom

A new construction method was created aimed to reduce the amount of labor exerted when constructing phantoms slice by slice while simultaneously providing CT dosimetry along the cranial caudal axis of the phantom. In addition, using a higher z-axis resolution provides more realistic anatomy, which was previously limited to 5 mm. The method utilizes constructing the phantom in components with the surfaces corresponding to dose locations relevant for obtaining dose information. Each section is constructed from a mold generated by a 3D printer. By building the phantoms in volume
sections less time is dedicated to treatment of individual slices, and allows for mass production and introduction of tissue equivalent materials. Although the dosimetry is limited to points fabricated by the phantom, with careful planning accurate dosimetry can be accomplished. With the development of a reference pregnant female series, the computational phantoms can be rendered rapidly for use in experimental dosimetry.

**Materials and Methods**

A series of tissue-equivalent substitutes were created to match the radiological parameters of tissues in the energy of diagnostics x-rays. These tissue substitutes include: adipose tissue (ATES), glandular tissue (GTES), and a new soft tissue-equivalent substitute (STES). All material substitutes were evaluated for three measurable parameters that best describe the low energy x-ray properties of tissues: density, half-value layer for CT beam spectra, and Hounsfield units. The techniques for evaluating each of these properties are described in detail below. Formalisms for calculating the variance or uncertainty of using these materials are derived to be used in sensitivity studies. Sensitivity studies are performed using Monte Carlo techniques to evaluate the effects of material densities on the accuracy of obtaining doses. Typically a parameter such as density is varied and simulations are run to determine sensitivity of doses to material density. Density is a good parameter because it correlates linearly with the attenuation and absorption of x-rays.

**Measurement Techniques**

**Hydrostatic weighting technique**

The density for each material was evaluated using hydrostatic weighting, a method used for measuring the mass density of a human body. The advantages of using this method are volume information of the material (body) is not required and its
real-world application of measuring body composition. The procedure is derived from Archimedes’ principle using three measured values: the weight of the body in air, the weight of the body completely immersed body, and the density of the auxiliary liquid. The instrumentation required to precisely measure the density is a high sensitive scale and an associated density kit (Mettler Toledo). The density of the body can be calculated using Equation 5-1.

\[ \rho = \frac{A}{A-B} (\rho_o - \rho_L) + \rho_L \]  

\[ (5-1) \]

The density of a solid is determined with the aid with the aid of a liquid whose density \( \rho_o \) is known. The solid is weighed in air (A) and then within the auxiliary liquid (B). The second measurement corresponds to the buoyancy force acting on the solid. The density \( \rho_L \) is the density of air (0.012 g/cm³) at standard air temperature and pressure. The scale is capable of measuring within 0.0001 grams. The density of air is a function of temperature and pressure, and the density of the water is also dependent on temperature. The temperature of the liquid water was measured using a thermometer, and using a look-up table for distilled water density tabulated with water temperature, the density could be determined. Using error propagation the associated error of this method is:

\[ \sigma^2_{pour} = \left( \frac{\partial \rho}{\partial A} \right)^2 \sigma_A^2 + \left( \frac{\partial \rho}{\partial B} \right)^2 \sigma_B^2 + \left( \frac{\partial \rho}{\partial \rho_o} \right)^2 \sigma_{\rho_o}^2 + \left( \frac{\partial \rho}{\partial \rho_L} \right)^2 \sigma_{\rho_L}^2 \]  

\[ (5-2) \]

Taking the derivative of each term from Equation 1-2 and substituting into Equation 1-3 gives:

\[ \sigma^2_{pour} = \frac{B^2}{(A-B)^4} \sigma_A^2 + \frac{A^2}{(A-B)^4} \sigma_B^2 + \sigma_{\rho_o}^2 \frac{A^2}{(A-B)^2} + \sigma_{\rho_L}^2 \left( 1 - \frac{A}{A-B} \right)^2 \]  

\[ (5-3) \]
The measurable errors in both A and B coincide with the scales resolution which equals 0.0001 grams. For the measurement in water 0.0008 was used as a conservative estimate. The error in liquid density is 0.0003 and the error in air density is about 0.00003.

The fluctuation in density from pouring of the material was also evaluated by measuring the density of the material at different locations and then added to the error in using the density kit to obtain a final value for the variation in density:

$$\sigma = \sqrt{\sigma^2_{\rho, \text{scale}} + \sigma^2_{\rho, \text{pour}}}$$  \hspace{1cm} (5-4)

**Hounsfield number**

The Hounsfield scale is an attenuation scale relative to water that associates a number to each voxel in an image acquired from a CT slice based on the relative attenuation of the material within that voxel. The CT number, or Hounsfield unit, is defined using Equation 5-5 for voxel attenuation coefficient $\mu_{\text{voxel}}$ and linear attenuation coefficient of water, $\mu_{\text{H}_2\text{O}}$.

$$\text{CT number} = 1000 \times \frac{\mu_{\text{voxel}} - \mu_{\text{H}_2\text{O}}}{\mu_{\text{H}_2\text{O}}}$$  \hspace{1cm} (5-5)

The Hounsfield units (HU) were evaluated using a Toshiba Aquilion One volumetric CT scanner (Toshiba America Medical Systems Inc., Tustin, CA) at standard imaging parameters with tube current modulated exposure control. The average HU was determined from the selected regions of interest using clusters of pixels with areas approximately 10 cm².

**Linear attenuation coefficient**

The linear attenuation coefficient ($\mu$) is a quantity that characterizes how easily a material can be penetrated by an x-ray beam. This quantity is dependent on the
incident beam energy or spectra, and the physical properties of the material. X-ray or photon attenuation is modeled as an exponential function, expressed below in Equation 5-6. The intensity, $X$, transmitted by a thickness $t$, from $X_0$ the initial intensity incident to the attenuating medium.

$$X = X_0 e^{-\mu t}$$

(5-6)

The linear attenuation coefficient was measured experimentally using a clinical CT scanner to provide a realistic CT energy spectrum. The beam was collimated into a narrow beam geometry using a custom lead plate to reduce scatter contribution. Several samples were fashioned into 1 cm thick slabs to be used for evaluating the linear attenuation coefficient. Each sample’s thickness was measured using a digital caliper (General, New York City, NY). The samples were then placed consecutively in the x-ray beam and exposure measurements were taken after the introduction of a new sample. A curve was generated plotting exposure in logarithmic scale as a function of material thickness. The attenuation coefficient was determined to be the slope of the line. The overall error from the attenuation coefficient determination is calculated using error propagation in Equation 5-6.

$$\sigma = \sqrt{\frac{1}{t^4} R^2 \sigma_{thick}^2 + \frac{1}{t^2} \left( \frac{\sigma_X^2}{X^2} + \frac{\sigma_{X_0}^2}{X_0^2} \right)}$$

(5-7)

The variance is calculated by rearranging the exponential attenuation, Equation 5-6, to solve for $\mu$, and applying error propagation. The term $t$ is the thickness of the slab, and $R$ is the natural log of the ratio of the unattenuated exposure, $X_0$, to the attenuated exposure $X$. Additional sources of error such as the scatter contribution and
beam geometry were not included as they would be reflected in the coefficient of
determination from the data.

In order to compare the measured attenuation coefficient to the theoretical, a
target half value layer was calculated for the scanner’s beam spectra. The target HVL
was evaluated using Monte Carlo techniques. The source term for the clinical CT
scanner, developed by Long and the ICRU-44 defined soft tissue, breast, and adipose
tissue were simulated to determine a target HVL.\textsuperscript{136} The HVL was measured replicating
the setup geometry of the experimentally measured HVL. The target HVL provided a
necessary benchmark as the HVL is highly dependent on the x-ray tube’s inherent
filtration which has shown to vary scanner to scanner.

For tissue substitutes that involve the STES material and polystyrene micro
beads the linear attenuation coefficient may be calculated by using the following
expression.

\[
\mu = \rho_{\text{adipose}} w_{\text{STES}} \left[ \frac{\mu}{\rho} \right]_{\text{STES}} + w_{\text{poly}} \left[ \frac{\mu}{\rho} \right]_{\text{poly}} \tag{5-8}
\]

Since the ATES and LTES mixture is a composition of STES and polystyrene micro
beads, the attenuation coefficient can be calculated from the mass fractions (w) of each
component and the desired density of the adipose tissue. The value for \((\mu/\rho)_{\text{poly}}\) was
evaluated using the effective energy of the CT beam and the linear attenuation
coefficients in the NIST database.\textsuperscript{108} In Attix, the weighting factors are expressed as (1-
g)\(w\), where g is the radiation yield.\textsuperscript{84} However, for CT x-ray energies, the value of g is
essentially zero for the STES and polystyrene so the equation was reduced to just the
weight fraction.
Materials

Soft tissue-equivalent substitute (STES)

A similar urethane-based STES was designed to eliminate the density fluctuations in the previous urethane-based compound; “PMC 121/30 Dry” (Smooth-On, Easton, PA) combined with the CaCO₃ powder (Fisher Scientific, Hanover Park, IL), 2.8% by weight. A new urethane-based resin “VytaFlex 40” (Smooth-on, Easton, PA) was used in conjunction with “URE-FIL 7” (Smooth-on, Easton, PA), a high density powder that dissolves well with urethane resins and improves overall stability. The powder also increases the overall effective Z of the material to better replicate the attenuation characteristics of soft tissue. The target density for human soft tissue is 1.04 g/cc. The accepted range of HU numbers for soft tissue varies between 10 and 40. The STES was fabricated by combining equal amounts of the VytaFlex 40 constituents, and adding the appropriate amount of URE-FIL 7 until the target density was achieved. An electric mixer was used to ensure a humongous mixture of the VytaFlex compounds and a uniform distribution of dissolved filler. The STES is allotted over 16 hours to cure and harden before measuring. After the target density was achieved, the compound was evaluated for attenuation and HU characteristics. Several slices of STES material, approximately 0.9 cm thickness, were created to measure the linear attenuation coefficient for a CT x-ray beam.

Adipose tissue-equivalent substitute (ATES)

Although this material was never unintended to be incorporated into the pregnant female phantom, a series of tissue composites were made to represent bodily tissues similar to soft tissue but lower in density. The ATES was derived from the STES, described above, and adding poly-fil polystyrene micro beads (Fairfield Processing,
Danbury, CT) to reduce the density. The STES density was modified to achieve a variety of densities corresponding to different breast tissue compositions and adipose tissue densities in the body. The homogenous breast and adipose tissues were modeled as the STES and adjusted to the desired density using the micro beads. A density curve was developed by adding various amounts of micro beads, by mass percent, and measuring the resulting density. Selective samples were further evaluated for their linear attenuation and HU properties using the techniques described in the above section. The development of these tissue-equivalent substitutes only adds more detail to the phantom and providing a more realistic physical model for obtaining accurate organ dose measurements.

**Bone tissue equivalent substitute (BTES)**

The BTES is an epoxy resin based material previously developed at UF. The material is a mixture of a fiber glass resin and Calcium and Silicon powders to obtain a similar chemical structure as human bone. The composite is formed 51.0% by mass of the fiber glass resin (Bondo, 3M™, St. Paul, MN) as well as 25.5% Silicon dioxide and 23.5% Calcium carbonate (Fisher Scientific, Hanover Park, IL). The BTES was developed as a homogenous bone mixture incorporating the cortical and trabecular spongiosa components seen in the human skeleton. The material was designed to match the density, mass attenuation coefficient, and mass energy absorption coefficients to Oak Ridge National Laboratory (ORNL) defined values within the diagnostic energy range. Accepted HU numbers for bone tissue show a wide range of values from 400 to 1000.
Phantom Construction Methodology

Design and development of anatomy

The pregnant female phantom series is constructed by Maynard from a set of computational phantoms derived from image sets of pregnant females who had undergone CT exams. Pregnant female anatomy was obtained from two sources of CT image sets: the Picture Archiving and Communication System (PACS) archive at the University of Florida, and publically anonymized images provided by Angel et al. All image sets were screened for coverage of the entire fetal area and no apparent physical abnormalities. Using the robust and flexible nature of NURBS/polygon mesh the contoured image sets were systemically combined with the UF adult non-pregnant computational phantom and four selected (2 week, 15 week, 25 week, and 38 week) UF reference fetal computational series. The 2 week and 15 week phantoms are very similar and can be combined to create a hybrid phantom segmented for the ovaries and uterus. These three phantoms were chosen from the eight phantoms in the library because they represented the significant gestation periods for peak radiobiological sensitivity for evaluating deterministic effects and cover the range of pregnancy for evaluating stochastic effects. The design criteria for the phantom library was adapted to generate reference phantoms with:

1. A single, reference orientation within the gravid uterus
2. Age-dependent target masses for the uterine wall, placenta, breast tissue, and total body mass
3. Preservation of original non-pregnant female reference tissue masses where appropriate, i.e. tissue not listed above
4. A single, reference maternal circumference, skin contour at each gestational stage
For completeness, a reference fetal orientation was desired to be maintained throughout the phantom series. The left occipital anterior (LOA) configuration was adopted as the reference fetal position as it is the ideal at-term configuration among the possible vertex presentations, which account for the vast majority of fetal presentations during birth.\textsuperscript{138, 139} Target organ, tissue, and whole body masses were derived from ICRP publication 89 and used as the basis for adjusting or fine-tuning organ size and sometimes location to accommodate the fetus and match target reference mass data, as well as fine adjustment of external anatomy. The pregnant female computational phantoms are illustrated in Figure 5-1.

**Selection of fetal anatomy for dosimetry**

Each phantom was built from a base phantom which contained the anterior anatomy and the pelvic bone and lumbar spine. The base phantom was derived from the 25-week phantom, containing no pertinent anatomy for dosimetry. The section of anatomy was further divided into components that each contributed a surface useful for dosimetry. A sagittal cut was done on each phantom that transverses the midline of the fetus. This cut split the fetus in half down the middle and provided the most area for dosimetry. Additional cuts relating to the head and body were performed in the axial plane to generate more dose locations for the 25 and 38 week phantoms. The number of cutting planes was limited to reduce the number of pieces and make 3D printing less complex. The selected cutting planes for each phantom are illustrated in Figure 5-2. The ovaries were selected to be segmented onto the 2/15-week phantom to evaluate the dose for the embryo. This is visualized in Figure 5-2a which includes an additional cutting plane to include the ovaries.
Fabrication using a 3D printer

The computational phantoms contain identification tags to label the regions of each organ and tissue when voxelized to a binary format. A software program uploaded each phantom and isolated the soft tissue and bone region of the phantom with those specific tags and converted each volume to a stereolithography file (.stl) which is the file format for 3D printing. This process had to be done because the original bone is made of complex polygon mesh surfaces that cannot be directly converted into .stl file format. After the soft tissue and bone structures of the abdomen/pelvic region were created they were modified to create a mold for casting. This was done using a robust computer-aided design program (Rhinoceros v5.0, Robert McNeel & Associates, Miami, FL) to extrude the surfaces into a new structure that encompassing the original surface.

The objective was to create a base phantom containing the lumbar spine and pelvis, and soft tissue not useful for fetal dosimetry containing no vital fetal anatomy, but the mother’s bone and muscle. The base phantom would provide structural support for the phantom when it would lay supine for dosimetry measurements during scanning. The first step was to render the pelvis and lumbar spine. The bone structure was divided into six molds: the lumbar spine, the sacrum, the left and right ilium, and the left and right ischium with the upper part of the femur. Each cast was built using a 3D printer containing the vacated cavity to be filled with the bone substitute. The bone substitute was introduced to each mold using heat and to decrease viscosity allowing the material to settle in the mold. Long thin rods were used to compact the material into small cavities and help the material settle. The bone substitute was allowed to cure over 24 hours, afterward it would be fully solidified. The open faces of the bone
structure were sanded, and then brushed with water to soften the mold. Using a screwdriver, the mold began to crumble when applied with pressure. The six bone equivalent structures were then reassembled using a strong adhesive (Krazy Glue, Columbus, OH) to form the complete pelvis and spine. The soft tissue component of the base phantom was built using 3D printer molds taken from the surrounding tissue of the pelvis and soft tissue posterior to the pelvis. The remaining soft tissue was fabricated using similar method. First the anatomy was sectioned based on the dose locations. Then each piece was further divided to meet printing specifications.

Results

Materials and Measurements

The STES density was optimized by combining 9.0% by weight of the URE_FIL 7 powder and 91.0 % Vytaflex 40, which resulted in a density of 1.041 g/cc. The STES was further evaluated for density fluctuations within a single pour by separating the sample into six individual samples. The results showed that the density varied among ten samples by 0.0026 g/cc. The total variation in density, calculated using Equation 5-4, was 0.00261 g/cc. The linear attenuation coefficient was calculated to be 0.228 cm\(^{-1}\) using linear regression. Applying Equation 1-8, the standard deviation in the measured attenuation coefficient is 0.0044 cm\(^{-1}\). The theoretical HVL calculation for the CT energy spectrum using the ICRU four-component soft tissue showed an effective attenuation coefficient of 0.229 cm\(^{-1}\) of soft tissue. The HU values for soft tissue were between 0 and 20 averaging around 10.

The ATES is the STES modified for density by adding the polystyrene microbeads to offset the density. Figure 5-4 displays the density function of the ATES vs. the amount of added micro beads. Error bars were calculated using Equation 5-4.
Fluctuations in density within a single sample were measured up to 0.0035 g/cc. ATES samples with densities of 0.98, 0.96, and 0.93 were selected to be appropriate for these bodily tissues: 50% (glandular) / 50% (fat) homogenous breast, 30% (glandular) / 70% (fat) homogenous breast, and subcutaneous adipose tissue respectively. From Equation 5-8 using the weight fractions from the fit to Figure 5-4 and the attenuation coefficient of the STES the attenuation coefficients were calculated. The effective mass attenuation coefficient of polystyrene was determined by using the effective energy of the CT spectrum.

The homogenized bone density was measured to be 1.60 g/cc with a variation of 0.05 g/cc. The half-value layer of the bone material was measured to be 1.7 mm at 120 kVp, and when simulated using the density of the BTES (1.60 g/cc) was determined to be 1.61 mm. From ICRU 46, the sacrum and lumbar column skeleton was listed to fall between 1.29 and 1.33 g/cc. This type of material is a homogenous bone tissue composition of the cortical and trabecular bone components. For comparison to other phantoms, the density of the Atom phantom is 1.60 g/cc and UF adult is 1.40 g/cc. The HU values fell between 650 and 810 with an average of 725.

**Completed Set of Phantoms**

All three pelvic/abdominal phantoms (2/15 hybrid, 25, and 35 week) are depicted in Figure 5-5. Figure 5-6 shows a topogram of the 15-week phantom. The phantoms were susceptible to air gaps where the pieces merged, and this was relatively unavoidable. However, it is easy to assemble and integrate with detectors. The color variation was due to using two different containers of one of the STES materials which varies in color among batches. The density did not fluctuate when switching containers. The mold was difficult to remove from the STES but relatively easy from the BTES. A
releasing agent (Smooth-On, Easton, PA) was used to aid the removing process but didn’t fully remedy the situation.

**Discussion**

The new STES soft tissue has the capability to be modified for reaching the target density by controlling the amount of URE_FIL 7 powder introduced to the mixture. The hydrostatic weighting technique is a proven methodology of verifying the target density of the tissue substitutes. As phantoms are being constructed samples should be collected and measured for density to maintain consistency. This was deemed important because different volumes of tissue being fabricated lead to variations in the density therefore the formula had to be modified to accommodate these deviations. Typically, this wasn’t an issue in the old construction process because phantoms were built in slices. However, now that the phantoms were able to be built in a manner that allows for the mass production of materials, a quality assurance protocol was necessary. With regard to the BTES it may be worth exploring a less viscous composite to ease the introduction of the substitute into the molds.

Generating molds from a 3D printer proved to be an accurate and automated method of fabricating phantoms. However, mold materials that were unsmooth resulted in difficulty in extracting the STES material from the mold. The BTES was easy to remove from the mold. The bone molds should try to incorporate as many open faces as possible to ease the material introduction process due to the tight cavities formed within the bone mold. The bone material is viscous and doesn’t pour well and has trouble settling into tight cavities without the aid of gravity or direct insertion. Therefore consideration and caution should be exercised when determining the size of each mold. Mass production of tissue components were a luxury in expediting the construction
process significantly. Pouring the entire soft tissue of all phantoms took “man hours” which is a significant improvement over the older method. In this situation the construction process is limited by how quickly the molds can be produced.

It was determined that making bone molds this way is too difficult. The soft tissue mold should be fabricated to surround the bone material by creating voids in the mold to be filled later. Most printers currently use plastic materials which have similar physical properties to human tissue. It may be beneficial to use the plastic as casing to assemble the mold and limit the air gaps. This was a big issue with the pregnant female phantoms. Mistakes to avoid in the future are to cut the phantom into components with orthogonal angles, use a more release friendly material, and build the soft tissue and bone mold as a single mold. With the flexibility of modifying the density of the BTES and STES, a two-component BTES could be created by increasing the density of the STES to represent the marrow cavities of human bone, and the BTES represent the cortical bone component.

**Conclusion**

This chapter explains a systematic approach of manufacturing computational phantoms for experimental dosimetry. This method has shown to streamline the production process of building phantoms by generating a smaller number of molds that allow for the mass production of tissue substitutes while offering a higher resolution of anatomy and provide customized dosimetry locations. In addition, the pregnant female physical phantoms are easy to assemble, produce high quality images, and are ideal for performing rapid dosimetry. The materials have been characterized to offer a range of tissue substitute options for incorporating more realism into each phantom.
Table 5-1. Commercially available anthropomorphic phantoms used for research

<table>
<thead>
<tr>
<th>Vendor</th>
<th>Name</th>
<th>Model</th>
<th>Genders</th>
<th>Cross Sectional</th>
<th>Cost</th>
</tr>
</thead>
<tbody>
<tr>
<td>Supertech</td>
<td>3D sectional torso phantom</td>
<td>CIRS 600</td>
<td>M</td>
<td>Yes</td>
<td>$17,664</td>
</tr>
<tr>
<td>Supertech</td>
<td>Virtually human male pelvis phantom*</td>
<td>CIRS 801-P-F</td>
<td>M</td>
<td>No</td>
<td>$19,156</td>
</tr>
<tr>
<td>The Phantom Laboratory</td>
<td>RANDO phantom</td>
<td>RAN100</td>
<td>M or F</td>
<td>Yes</td>
<td>$27,442</td>
</tr>
<tr>
<td>CIRS</td>
<td>ATOM® Dosimetry Verification Phantoms</td>
<td>702-D</td>
<td>M or F</td>
<td>Yes</td>
<td>$26,790</td>
</tr>
</tbody>
</table>

* organs includes: bone, cartilage, spinal cord, vertebral disks, muscles, intestines, bladder, prostate, rectum, and interstitial fat

** all prices do not include plug ins for dosimeters
Figure 5-1. Frontal view of the pregnant computational hybrid phantom’s pelvis. From left to right, a) 15, b) 25, and c) 38 week computational hybrid phantoms.
Figure 5-2. Selection of cutting planes to divide the abdomen/pelvis of each phantom. Ovaries are illustrated in (a) as well as all of the fetal anatomy in all phantoms. Cutting planes are in orange. From left to right, a) 15, b) 25, and c) 38 week computational hybrid phantoms.
Figure 5-3. Progression of mold fabrication. First the issue of interest is isolated (a) in a CAD software program. Using a series of modeling tools a computational mold is created with a vacated volume equal to the original object (b). The mold is then rendered using a 3D printer (c). (Photo courtesy of Matthew R. Hoerner)
Figure 5-4. Density curve for ATES based on added weight percentage of micobeads. Linear regression applied to determine the best fit equation.

Density = \(-0.238(\% \text{ added}) + 1.007\) g/cc
Regression coefficient: 0.9914
Figure 5-5. Pregnant female anthropomorphic abdomen/pelvis phantom series. The three phantoms are a) 2/15 week hybrid b) 25 week and c) 38 week. Photo courtesy of Matthew R. Hoerner.
Figure 5-6. Topogram of the 25-week phantom (below the orange line) and the thoracic region of the adult reference female physical phantom (above the orange line). White lines at the top are from the metal cords used to bind the phantom together.
CHAPTER 6
MEASURED CONCEPTUS DOSES ON AN ANTHROPOMORPHIC PHANTOM FROM MULTI-DETECTOR COMPUTED TOMOGRAPHY STUDIES USING TUBE CURRENT MODULATION FOR ALL GESTATIONAL STAGES

Introduction

As highlighted in Chapter 1, numerous studies have been performed to evaluate radiation absorbed dose to an unborn child from multidetector CT (MDCT). Some of these studies have employed Monte Carlo methods while others performed physical dosimetry utilizing anthropomorphic phantoms. In the case of Monte Carlo, none of those investigations were able to model a source term capable of evaluating absorbed fetal dose from CT studies using tube current modulation (TCM). This research is currently being done making computational pregnant female dosimetry certainly possible of being done in the near future.\textsuperscript{136} The physical dosimetry that has been performed, however, has never incorporated anatomical models portraying a pregnant female in the mid to late stages of pregnancy. Most have been relegated to using reference female phantoms. By using adult female or male reference phantoms these studies are restricted to measuring uterine radiation doses, because none of the phantoms have delineated locations of the developed fetus. In addition, because of individual differences in anatomy the radiation dose in the first trimester will be different than in the second, and third. This clear lack of anatomical phantoms representing the second and third trimester prompted the investigation into radiation exposures from CT with tube current modulation techniques.

Exposures from CT radiation delivered to a child-bearing mother are limited to few cases. These cases include but are not limited to: pulmonary embolism (PE), acute appendicitis (AA), and trauma (T) studies. These cases represent the overall majority of
situations where the fetus receives direct exposure from radiation in CT. Other cases which use CT for diagnosis such as pancreatitis and Crohn’s disease are not associated with pregnancy.\textsuperscript{140} AA occurs at a rate of 0.07-0.18% during pregnancy.\textsuperscript{141, 142} Ultrasound is the preferred imaging method during the 8\textsuperscript{th} to the 15\textsuperscript{th} week post-conception due to sensitivity of the fetus to teratogenetic effects. However, ultrasound is very user-dependent and the third trimester is more difficult to perform due to the enlarged uterus and displacement of the appendix. Maternal perimeter is certainly a factor in determining the imaging approach. For a patient in an advanced stage of pregnancy, where the abdomen and pelvis have enlarged, CT provides the most information.\textsuperscript{143} Magnetic resonance imaging (MRI) can provide another option to avoid fetal exposures. MRI has shown to produce comparable images for diagnosis of acute appendicitis; however, it is not the preferred method of obtaining a diagnostic image.\textsuperscript{144, 145} PE most commonly results from a blood clot in the deep veins of the legs or pelvis. It plays a significant role in maternal mortality in the United States. One study has shown that 40% of maternal deaths are related to pulmonary embolus or cardiac disease.\textsuperscript{146} Incidence occurs between 0.05% to 0.3% of all pregnancies and is diagnosed through the use of CT angiography (CTA).\textsuperscript{147} Pregnancy complications such as hemorrhaging of the uterus, placental abruption, placenta previa, uterine bleeding, disseminated intravascular coagulopathy are leading causes of maternal mortality in the United States and require immediate medical attention.\textsuperscript{146}

Clinics will often revise protocols to limit excessive radiation and preserve sufficient image quality to obtain a diagnosis. These dose reduction methods include: lowering the tube current directly or indirectly by increasing the noise factor in TCM
studies, shortening the scan length to cover only the critical anatomy, and increasing the pitch. Modern volumetric scanners have the ability to capture entire organs and volumes of anatomy in a single rotation. These next generation scanners produce images with more coverage of anatomy in less amount of time offering new advanced procedure possibilities in non-invasive diagnostic imaging. Currently the four major vendors (Toshiba, Siemens, General Electric, and Phillips) offer scanners with at least 64 channel (slice) configurations. Volume scanners are dynamic in their ability to perform studies at a variety of beam widths and slice thicknesses.

TCM techniques have become the standard imaging protocol for pregnant women undergoing CT scans. TCM protocols are designed to optimize dose and image quality to generate the best images possible without using excessive radiation. Dose and image quality are viewed as being optimized since the exposure parameter allows a tradeoff between image quality and exposure. For abdominal/pelvis protocol scans the beam intensity profile is relatively consistent over the scan due to little variation in tissue composition and effective diameter of the patient between imaging slices. For other exams, the chest protocol for example, TCM greatly affects the x-ray dose distribution within the patient. The tube current (mA) ramps up around the shoulders to compensate for the amount of bone in the shoulders then decreases over the lungs (which have 33% the density of soft tissue). The second and third trimester pregnant female patient produces a more dynamic mA(z) profile with more deviation than the first trimester mother. The circumference of the abdomen and pelvis are dependent on position along the z-axis due to the protrusion of the mother's belly.
The absorbed radiation dose to the fetus was investigated from a series of TCM protocols using a volumetric scanner with variable image acquisition techniques to provide dosimetry information from the most modern scanners. Furthermore, the absorbed doses from among all gestational stages were used to assess the effect of TCM and acquisition settings.

**Materials and Methods**

Chapters 3 through 5 have detailed the experimental tools and measurement techniques to rapidly and accurately quantify radiation absorbed dose to the unborn child in an adult reference pregnant female. In Chapters 3 and 4, two types of dosimeters were investigated for their performance in Computed Tomography (CT) through comprehensive characterization methods. From these characterizations, these dosimeters are on the cutting edge of CT experimental dosimetry. It is with this goal that they may be used in future studies to familiarize the radiological measurement physics community with their capabilities. In Chapter 5 a novel anthropomorphic phantom was constructed to be used with these dosimeters to provide physical anatomy necessary to evaluate radiation absorbed dose to a pregnant woman spanning the period of gestational stages. The integration of these tools make it possible to collect data and evaluate the fetal exposure from CT in pregnant female studies.

**Anthropomorphic Phantoms**

The UF adult female along with the set of four abdominal/pelvis pregnant female reference phantoms were used in this study to evaluate the conceptus dose along with a set of fiber-optic dosimeters. Each phantom represents an adult pregnant female with gestational ages of 2/15 week, 25, and 35 weeks and based on the pregnant female hybrid computational model. Each phantom represents the 50th percentile female for a
general population by targeting ICRP 89 reference organ mass, height, weight, and weight gain for each stage in pregnancy. A more detailed description of each phantom can be found in Chapter 5.

Measurement Techniques

In order to evaluate the absorbed dose to the fetus, a series of linear fiber-optic dosimeters (LFODs) were used with the pregnant female physical phantom series. LFODs are valuable tools for measuring CT organ doses since they have an ability to acquire linear dose measurements which are more realistically represent the geometry of the target volume of interest than do conventional point dosimeters. The LFOD dosimeters were calibrated from a series of measurements within a cylindrical PMMA phantom assembly. Figure 6-1 illustrates the side view of the cylindrical phantom with the holes utilized for calibration and the positions of each detector relative to the primary x-ray beam. The detectors were calibrated by exposing the dosimeters to a helical scan over a 40-cm scan length to reach dose equilibrium.

In all, 2 positions were evaluated corresponding to the top and center holes. The reference detector, a 0.6 cc small volume ionization chamber (Radcal®, Monrovia, CA) evaluated the air kerma at each position from a helical scan with pitch equal to 1.5. Two sets of count data were generated. One set corresponds to the primary beam conversion coefficient, while the other corresponds to the scatter radiation. Using the count rate data from the scan acquisition a scatter and primary coefficient was determined to find the number of detected counts originating from scattered radiation and primary radiation. The data was optimized iteratively to find the correction factor that relates measured signal to air kerma. For each air kerma measured, \( K \), \( C^{Sig} \), counts were recorded from the plastic scintillator and \( C^R \) counts from the reference fiber.
To find the primary and scatter conversion coefficients, $\lambda^P$ and $\lambda^S$, respectively as well as $\lambda^R$, the data was entered into a custom MatLab program that used a "\" technique to optimize the conversion coefficients by modeling the data from Equation 6-1.

$$K_1 = \frac{\lambda^P}{\lambda^R} \left[ \frac{C_{i1}^{Sig,P} - \lambda^R C_{i1}^{Ref,P}}{C_{i1}^{Sig,P} - \lambda^R C_{i1}^{Ref,P}} \right] + \frac{\lambda^S}{\lambda^S} \left[ \frac{C_{i1}^{Sig,S} - \lambda^R C_{i1}^{Ref,S}}{C_{i1}^{Sig,S} - \lambda^R C_{i1}^{Ref,S}} \right]$$

(6-1)

where the indexes 1 and 2 correspond to the hole location within the phantom. This calibration should be performed separately for each tube voltage. The signal and reference counts ($C_i^{Sig,H}$ and $C_i^{Ref,H}$) are summed for each integration time above the counts originating from the PMT.

**PMT Control Program**

The PSD dosimetry system contains an array of five PMTs permitting both dosimeters to be monitored simultaneously. A control program was developed in the form of a GUI (MatLab, Mathworks, Natick, MA) to control and analyze data from both dosimeters simultaneously, and include input boxes for all necessary parameters to correct the light output and produce immediate dose measurements. These parameters include: integration time, dead time, background correction factor, energy correction factor, and scatter and primary calibration factors. All of these parameters have been discussed in previous chapters and are highlighted in the conclusions section of Chapters 3 and 4 how they can be incorporated into a dosimetry system. The completed GUI is illustrated in Figure 6-2 to display all user functions and numerical inputs and outputs. Each detector requires a sampling time for all PMTs it occupies, a reference fiber correction factor for the plastic scintillator, and dead time values for all scintillators and reference fibers if desired. The program produces the measured air kerma values for each dosimeter. The GUI script code is attached in Appendix A.
CT Procedures

The anthropomorphic phantoms were imaged with a Siemens Sensation 16 scanner (Siemens Healthcare USA, Malvern, PA) using tube current modulation. The scanning parameters are listed for variations of routine abdomen/pelvis (appendix), CTA chest pulmonary artery (PE), and body trauma scanning protocols from UF Health radiology. The anatomical landmarks defining the beginning and end of each exam are illustrated. The PE acquisition begins at the base of the adrenals and ends above the apex of the lung. The CT body trauma scan begins at the top of the kidneys or the dome of the diaphragm and terminates at the lesser trochanter. The appendix protocol spans the length of the diaphragm to the lesser trochanter.

Assessing Dose to the Fetus

The LFOD measures the air kerma along the axis of the fetus’s midline. To obtain more realistic dosimetry measurements another LFOD was implemented to measure the dose in the horizontal plane. One was used for the head, and the other for the chest and abdomen resulting in three air kerma measurements. To find a single fetal dose, the values were scaled based on the length of the each dosimeter. This is derived in Appendix B which is based on weighting each fiber by its active length.

Once the total body air kerma dose was measured, the radiation absorbed dose to the fetus was calculated. This was done by weighting the effective mass energy-absorption coefficients of soft tissue and bone component of the fetus by mass.

\[
D_{fetus} = K_{fetus} \left[ w_{bone} \frac{\mu_{en}}{\rho_{air}} \right] + w_{tissue} \frac{\mu_{en}}{\rho_{air}} \]  \hspace{1cm} (6-2)

\(K_{fetus}\) is the measured air kerma from the detector and \(w\) is the weight fraction.

This is similar to the method proposed by Attix of weighting chemical compounds by
element. Most tissues are homogenous in nature and then typically be assigned a corresponding correction for material difference. The high magnitude of x-ray energy-absorption characteristics in bone leading to higher doses in CT required consideration. It was assumed that the bone and soft tissue are mixed together into a single entity and that the average air kerma is applicable to both. Other studies simply just use an f factor that converts air kerma to soft tissue dose. The effective mass energy-absorption coefficients were evaluated by using the normalized energy spectrums measured in Chapter 3 for air and polystyrene. The bone and soft tissue spectrum were equated to the polystyrene spectrums. The values are relatively independent of depth as demonstrated in Chapter 3.

**Results**

**Absorbed Dose Measurements**

Using the reference values for the weight of bone and soft tissue for each gestational period, the dose factor was calculated using Equation 6-2 to find the increase in radiation absorbed dose when accounting for the amount of homogenous bone in the fetus. Equation 6-3 was evaluated using mass energy absorption coefficients at 50 keV for bone, soft tissue and air. The calculated values are tabulated in Table 6-1. Note that in the early gestational periods the fetus is comprised mostly of soft tissue, but at the later stages of pregnancy the bone begins to fully develop representing over 10% of the total fetal mass.

Scans were performed using the default scan parameters for a Chest Abdomen Pelvis (CAP) exam and modifying the scan length to include the anatomy described in the above section. Table 6-2 shows the two protocol settings. One setting the beam energy was set to 120 kV and the other was 140 kV. For this study the tube current
was varied by controlling the preset tube-current product value which directly monitored the noise level of the exam. A higher setting would result in more signal and less noise.

Tables 6-3 through 6-6 show the measured fetal/embryo absorbed doses for each of the gestational physical phantoms. The radiation absorbed dose to the 2-week embryo was evaluated by taking measurements at the position of the ovaries in the 2/15-week hybrid phantom. The average fetal dose for the 15th week fetus was 6.84 mGy per 100 mAs with a maximum fetal dose of 8.67 mGy per 100 mAs and a minimum of 5.45. The range of values is well below 100 mGy, which is the established dose threshold that may induce teratogenic effects. Dose measurements were normalized by the effective tube-current delivered to phantom from each exposure as reported by the scanner. The calculated CTDIvol was adjusted based on the effective tube current. For the 120 kVp scans the average of the CTDIvol was 0.07026 mGy/mAs and 0.1022 mGy/mAs for 140 kVp. From the collected data, little difference in dose was seen between the CAP, T, and AA exams. The exception was for the 38 week fetus which is larger and located more superior causing it to receive more scatter radiation. The PE exam was completely out of the scan field and resulted in small amounts of accumulated dose.

**Dose Estimation Models**

Figure 6-3 plotted the relationship between CTDIvol and fetal dose. Curves are linear but display different fitting functions. Each curve contains data from both energies and mA settings. For high scanner outputs the values begin to deviate from one another and a single model becomes difficult. Each data set was fit to a first-degree polynomial functions. The fit for the 25-week phantom is $y = 1.0554x - 0.71948$, and the fit for the 15-week phantom is $y = 0.8465x + 0.11707$. The linear no-threshold
fit is 1.0019 mGy/mGy ($R^2 = 0.9921$) and 0.8561 mGy/mGy ($R^2 = 0.9820$) for the 25 and 15 week phantoms respectively. This is more of an ideal fit since it has physical meaning since if the tube current was set to 0 then the CTDI would be 0 and so would the fetal dose.

Another interesting relationship that is often not documented is the relationship between dose and scan length. Since data is acquired by the detection system in real time and is recorded a count vs. time profile can be generated. Since the x-ray tube moves along the z-axis, counts vs. position plot can be formulated to show accumulated dose vs. scan length. Figure 6-4 shows the total radiation absorbed dose as a function of scan length (moving inferior to superior and vice versa) for the 15 and 25 week phantoms at 120 kV. The data shows that the developed fetus is more sensitive to changes in scan length which can be attributed to more scatter being generated. Before 200 mm on a CAP exam (head first) no dose is received by the fetus. This Figure may help clinics set protocols for PE exams which mainly focus on the chest through the liver (pulmonary artery).

Discussion

This study presented unique experimental methods of evaluating fetal radiation absorbed dose to multiple gestations from tube current modulated scanner techniques, also showing that caution should be exercised when using the CTDI size-specific correction factors found in literature. The conversion coefficient between absorbed dose and CTDIvol for the 25-week phantom was higher in magnitude than the 15-week phantom. This contradicts the AAPM report that uses size-specific correction factors to correlate CTDIvol and absorbed dose, which found that a larger perimeter resulted in a lower dose due to attenuation. However, a significant factor in the increase in absorbed
dose is the increase of the weight fraction of bone within the fetus. When using TCM the effects of patient size are minimized because the x-ray fluence attenuated by the patient will be compensated by the x-ray tube to maintain a consistent signal level hitting detector array. Also, CTDIvol reported for this scanner is based on the average mA not the mA irradiating the fetus which is probably the maximum mA given that the mother’s waist is at the largest circumference at the fetus’s z-axis position. Another interesting discovery was that at older gestational ages are more sensitive to changes in the mA setting. The 15-week phantom absorbed radiation dose is approximately 0.047 mGy per mAs at 120 kV, compared to the 0.082 mGy per mAs of the 25-week phantom at 140 kV from a CAP exam. This is mostly likely due to the scatter radiation contribution near the fetus.

The advantage of using LFOD’s and these anthropomorphic phantoms is evaluating organs located on the periphery on a big cross section. The radiation dose is very dependent on the angle of the tube as it transverses over the organ. At anterior angles lower doses will be obtained and posterior angles will produce higher doses. Point dosimeters placed in a cross sectional phantom will only monitor the dose for that single plane. If the organ is of a length that the tube will make it back around those doses will be not captured and the dose will be inaccurate. If multiple rotations are made around the organ than the dose readings will average out and approach the actual organ dose. These circumstances are dependent on the exam pitch, organ length, and slice thickness. Additionally, if the dosimeters are placed at equal distances apart in the z-axis then there is a probability that the beam will intersect each dosimeter in the same location and the true organ dose value will never be captured. These
issues would never be significant using a LFOD placed along the organs length in the z-axis because measurements are obtained along the entire length.

The calibration methods being implemented by the literature discussing CT dosimetry are subject to scrutiny. Free-in-air calibrations will not account for the scatter radiation exposed to the detector. All three studies that performed physical measurements calibrated their TLDs free-in-air. Scatter radiation is a different energy than primary x-ray radiation and will have a different correction factor. Most of the fetal dose measurements were obtained in the central locations of the phantom, where scatter is a significant contribution to the total absorbed dose.

One of the limitations to this study was obtaining data from a phantom simulating a single body habitus with a conceptus oriented in a fixed position. Radiation doses vary depending on the size, fat/muscle composition of the phantom, and the position of the fetus within the mother. Preliminary sensitivity studies using Monte Carlo have placed the uncertainty in measuring doses from varying soft tissue density to be anywhere between 0.04 to 0.06 mGy per 0.01 g/cc. However, this value was calculated from simulations performed on an adult reference male. Therefore a soft tissue density of 1.02 g/cc would result in an additional radiation absorbed dose of 0.12 mGy (1% difference). It would be expected that the larger pelvis diameter would have a higher magnitude of an effect. Exact doses will depend on the imaging protocols established by the institution.

Imaging techniques to lower the absorbed dose that are applicable to MDCT include but are not limited to: lowering the tube current, minimizing the scan length to reduce scatter contributions, and increasing the pitch. Increasing the pitch and lowering
the tube current may result in poorer image quality. Modifying the scan length would have a greater effect on regions in the center of the patient which receive more dose contribution from scatter radiation. Changing the pitch may or may not have an effect on regions located on the periphery because the primary beam may still directly pass through that region. This is especially true for smaller organs where the primary beam transverses the plane of the region in a single pass, which are more likely to occur at wider slice thicknesses.

A potential technique to lower the fetal dose in MDCT that is applicable to volumetric scanning would be to raise the pitch and optimize the starting angle to control the tube position. This would create a scenario where the direct x-ray beam passes through the fetus mostly from the posterior surface of the mother’s skin and less from the anterior surface. Although currently this cannot be done since the starting angle is uncontrollable, it poses as interesting and potential technique to monitor exposures in CT.

**Conclusion**

Radiation doses to the embryo and fetus from the appendix and body trauma TCM scans are shown to contribute a relatively equal amount of dose that falls below the level of concern. It was expected that the PE protocol would yield lower radiation doses than the other protocols because the fetus/embryo is only exposed to scatter radiation. The fetal dose from a PE exam is most dependent on the scan length in the inferior direction of the patient. The basis of these dose estimates show that the doses are highly dependent on the exam protocols especially the tube potential. Clinics that use 140 kV tube voltage and high tube current settings will irradiate the fetus at a much higher level than clinics using low-dose protocols. Some of the results of this study are
applicable to large patient dosimetry and could result in optimizing exam protocols to minimize excessive dose. Finally, this study can potentially serve as a basis to develop a clinically practiced fetal dose calculation methodology from its empirical finds as well as its potential to validate Monte Carlo TCM simulations for various CT scanners.
Table 6-1. Dose coefficients for converting fetal soft-tissue doses to total body doses.

<table>
<thead>
<tr>
<th>Gestation</th>
<th>Bone mass (g)</th>
<th>Total fetal mass (g)</th>
<th>$\mu_{en}/\rho$ ratio of bone to soft tissue</th>
<th>Total body dose coefficient</th>
</tr>
</thead>
<tbody>
<tr>
<td>15 week</td>
<td>&lt; 1</td>
<td>160</td>
<td>1.105</td>
<td>1.00</td>
</tr>
<tr>
<td>25 week</td>
<td>56.85</td>
<td>990</td>
<td>1.567</td>
<td>1.09</td>
</tr>
<tr>
<td>35 week</td>
<td>293.06</td>
<td>2700</td>
<td>2.486</td>
<td>1.27</td>
</tr>
</tbody>
</table>

Table 6-2. Imaging parameters for 16-MDCT scanner.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Protocol A</th>
<th>Protocol B</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak kilovoltage (kV)</td>
<td>120</td>
<td>140</td>
</tr>
<tr>
<td>Detector configuration</td>
<td>16 x 1.5 mm</td>
<td>16 x 1.5 mm</td>
</tr>
<tr>
<td>Pitch</td>
<td>0.75:1</td>
<td>0.75:1</td>
</tr>
<tr>
<td>Tube rotation (sec)</td>
<td>0.5</td>
<td>0.5</td>
</tr>
<tr>
<td>Normalized CTDIvol (mGy/mAs)</td>
<td>0.0703</td>
<td>0.102</td>
</tr>
<tr>
<td>Avg. scan length: CAP (mm)</td>
<td>661</td>
<td>661</td>
</tr>
<tr>
<td>Avg. scan length: AA (mm)</td>
<td>360</td>
<td>360</td>
</tr>
<tr>
<td>Avg. scan length: T (mm)</td>
<td>290</td>
<td>290</td>
</tr>
<tr>
<td>Avg. scan length: PE (mm)</td>
<td>330</td>
<td>330</td>
</tr>
</tbody>
</table>
Table 6.3. Radiation absorbed embryo doses from a variety of pelvic exams.

<table>
<thead>
<tr>
<th>TCM setting (mAs)</th>
<th>Radiation absorbed fetal dose (mGy)</th>
<th>Avg. T/AA/CAP absorbed dose [mGy]</th>
</tr>
</thead>
<tbody>
<tr>
<td>280</td>
<td>10.89</td>
<td>13.68</td>
</tr>
<tr>
<td>240</td>
<td>9.16</td>
<td>11.97</td>
</tr>
<tr>
<td>200</td>
<td>8.21</td>
<td>10.66</td>
</tr>
<tr>
<td>160</td>
<td>6.52</td>
<td>8.74</td>
</tr>
</tbody>
</table>

Table 6.4. Radiation absorbed 15-week fetal doses from a variety of pelvic exams.

<table>
<thead>
<tr>
<th>TCM setting (mAs)</th>
<th>Radiation absorbed fetal dose (mGy)</th>
<th>Avg. T/AA/CAP absorbed dose [mGy]</th>
</tr>
</thead>
<tbody>
<tr>
<td>280</td>
<td>12.30</td>
<td>15.47</td>
</tr>
<tr>
<td>240</td>
<td>9.61</td>
<td>13.21</td>
</tr>
<tr>
<td>200</td>
<td>8.20</td>
<td>11.24</td>
</tr>
<tr>
<td>160</td>
<td>6.50</td>
<td>8.83</td>
</tr>
</tbody>
</table>

180
Table 6-5. Radiation absorbed 25-week fetal doses from a variety of pelvic exams.

<table>
<thead>
<tr>
<th>TCM setting (mAs)</th>
<th>Radiation absorbed fetal dose (mgY)</th>
<th>Avg. T/AA/CAP absorbed dose [mgY]</th>
</tr>
</thead>
<tbody>
<tr>
<td>280</td>
<td>16.98</td>
<td>20.85</td>
</tr>
<tr>
<td>240</td>
<td>14.45</td>
<td>15.78</td>
</tr>
<tr>
<td>200</td>
<td>11.76</td>
<td>12.48</td>
</tr>
</tbody>
</table>

Table 6-6. Radiation absorbed 38-week fetal doses from a variety of pelvic exams.

<table>
<thead>
<tr>
<th>TCM setting (mAs)</th>
<th>Radiation absorbed fetal dose (mgY)</th>
<th>Avg. T/AA/CAP absorbed dose [mgY]</th>
</tr>
</thead>
<tbody>
<tr>
<td>280</td>
<td>19.03</td>
<td>27.88</td>
</tr>
<tr>
<td>240</td>
<td>16.09</td>
<td>22.98</td>
</tr>
<tr>
<td>200</td>
<td>13.51</td>
<td>19.18</td>
</tr>
<tr>
<td>160</td>
<td>10.80</td>
<td>15.11</td>
</tr>
</tbody>
</table>
Figure 6-1. Phantom assembly used to calibrate the dosimeters. Two CTDI 32-cm body phantoms and three soft tissue-equivalent phantoms provided over 40 cm of build up to reach dose equilibrium along the central axis. (Photo courtesy of Matthew R. Hoerner)
Figure 6-2. MATLAB GUI dosimetry program with user functions and output data.
Figure 6-3. Relationship between fetal dose and CTD\text{vol} at various gestational ages for a CAP examination.
Figure 6-4. Accumulated fetal dose as function of scan position. Data taken from the CAP scans (head first) using protocol A.
CHAPTER 7
CONCLUSIONS AND FUTURE WORK

Plastic Fiber-Optic Detectors

General Notes

The effectiveness of PSDs as a dosimetry tool is dependent on many parameters such as the particle energy, exposure rate, photon detector, and exposure of the optical fiber. From a diagnostic energy perspective the effect of quenching is very apparent and it seems it may be more productive to explore other types of scintillation materials. This is especially pertinent to the development of a dual-scintillating dosimeter which would be more highly effective using scintillators that are not prone to quenching in order to better resolve the beam energy spectrum and make appropriate corrections. This will be discussed in further detail in the future work section. When using these detectors for megavoltage x-ray and particle beams a noticeable field size and dose rate is apparent. However it was demonstrated to be correctable, though not ideal for performing accurate dosimetry. When looking at other studies it appears that these issues stem from the photon detector’s time response from pulse counting compared to signal integration. This would suggest moving towards a more conventional photon detector, such as a CCD camera that integrates the signal. This work is currently being done at University of Florida for Cone-Beam CT (CBCT) dosimetry applications.

This dissertation has clearly demonstrated the correction methods for count rate effects stemming from dead time for organic and inorganic scintillators using a photomultiplier tube array. Using the correction method detailed in this document provides a universal technique for characterizing PSDs for any type of radiation detection and measurement application. The direct relationship between dead time and
Cerenkov radiation is not definite but seems very likely. Our analysis demonstrates that the irradiation of an optical fiber, and possibly scintillator, create light attenuation/absorption issues that may be more or less magnified depending on the type of photon detector used. When transitioning from the PMT to the CCD, the count rate effects and stem effects from high energy radiation should be characterized and applied directly to the dead time model in this manuscript. It should be noted however that the nonparalzable model is an ideal and not applicable to all types of detectors.

Linear fiber-optic dosimeters offer the most utility for performing organ dose measurements in the radiological energy range owing to their spatially integrated dose measurement. The application of one dimensional dosimetry is relatively new and presents more opportunities to expand the capabilities of these devices into more medical modalities. In addition, the real-time monitoring of counts from helical scanning has shown promise in reducing the energy dependence of a detector by measuring the scatter and primary x-ray beam contribution. The limited cases presented in this study show that identifying the scatter and primary radiation contribution can self-correct the detector for energy and depth dependence to perform accurate dosimetry.

Energy Dependence of a Plastic Scintillator

The results obtained in this dissertation have produced a general notion to the response of a plastic scintillator to irradiation from radiological x-rays. Figure 8-1 illustrates the mass energy absorption coefficients of polystyrene to air for energies ranging from 1 to 150 keV. A linear fit ($R^2 = 0.9986$) was applied to the data between 35 and 75 keV to demonstrate the linearity of the energy dependence between those energies. For diagnostic x-ray beams characterized by an effective energy between 35 keV and 75
keV, any change to that spectrum would increase or decrease the energy dependence of the detector proportional to a constant. The linearity of the energy dependence was characterized in this manuscript as well as Hyer et al. The energy dependence of a plastic scintillator in CT can be described empirically as a function of both beam energy and depth. Equation 8-1 formulates an expression for the correction factor of a plastic scintillator to appropriately correct for the energy dependence.

\[
C = \beta_p (\alpha_1(E) * d + \alpha_2 E) + \beta_s C_s
\]  

(7-1)

Where \(\beta_p\) and \(\beta_s\) are the fraction of counts that are a result of interactions from the primary radiation and scatter radiation with the scintillator. \(E\) is defined as the effective energy of the x-ray beam free-in-air. The primary beam coefficients (\(\alpha_1\) and \(\alpha_2\)) are model the energy dependence coefficient from changes to tube potential and depth. Equation 8-1 could serve as a model for correcting the response of the detector to the response of an ionization chamber.

These investigations have shown that plastic scintillators used in radiological dosimetry demonstrate a quenching effect. Further evidence of quenching is supported by the slope of Figure 8-1 compared to the slope of the plastic scintillating dosimeter. From Figure 4-6 the slope of that line when converting from kVp to effective energy of each spectrum is 0.021 per keV. This value is double the 0.0112 per keV value from applying large cavity theory. This may explain why Lessard et al. was able to use large cavity theory to correct for the response of the detector without accounting for quenching. The quenching parameter can be hidden within the scintillating efficiency or any other conversion parameter in the 60-140 kVp energy range. The work of Frenlin et al. showed that quenching is linear with respect to energy in this energy range. When
accounting for quenching and mass energy absorption, the sensitive element
demonstrates a linear energy dependence from direct radiation in CT. When scatter is
present this linearity breaks down as evidence by the results in Chapter 3, and the right
term in Equation 8-1 dominates the energy dependence. The magnitude of the
quenching effect observed by the BCF-12 plastic scintillators used in our detectors
appears to be significantly higher than those of other studies. The cause of this is
largely unknown.

Anthropomorphic Phantoms
There is a great benefit of rendering phantom molds using 3D printers compared to the
traditional method using a 2D engraver. However, using a 3D printer brings a higher
cost associated with building each mold. In addition constructing molds is complicated
by the limiting build volume of a 3D printer (10” x 8” x 8”) compared to the surface area
of a 2D engraver (24” x 16”). Building anthropomorphic phantoms means the full cross
sections of adults must be composed of multiple components, which adds to the build
cost and construction time. Considerations must include the ease of extracting the soft
tissue material from the mold and the marginal cost associated with each mold. For this
project 275.55 in³ of material was used at an estimated cost of $826.65 compared to the
approximate $100 of foam board that would be required for this project. However, more
tissue material would have been needed because each phantom would have been built
separately instead of into components which helps alleviate the overall cost of the
project.

Physical Dosimetry

One of the findings that these investigations have founded is that scatter
radiation plays a prominent role in the radiation absorbed dose received from CT
exposures. This scatter radiation is a completely different energy and induces a completely different detector response than primary radiation. When using a detector with a sensitive element that is greater than that of air or soft tissue its correction factor will be less than 1. At low energies the correction factor will be the smallest and at higher energies it will start to approach 1. If the detector is calibrated free-in-air it will overestimate the correction factor because the scatter contribution, which has a lower coefficient, is ignored. In this case CT doses are overestimated. When using a detector with a smaller effective Z, the reverse case is true and radiation absorbed doses will be underestimated. The calibration method presented in this study is ideal for CT dosimetry because it will properly simulate the scatter and primary radiation experienced by each detector. Using a CTDI phantom and a single rotation non table translation will be a closer approximation, as it produces an estimated 2/3 primary and 1/3 scatter. However, for large patients the actual value for a CT scan approaches equal contributions of both. In any case, it is a much better estimation than free-in-air calibration methods.

A review of the detectors described in Chapter 3 used the dosimetry studies presented are LiF TLDs (TLD-100, Harshaw, Solon, OH) and MOSFET (1002RD, Best Medical Canada, Ottawa, Ontario) dosimeters. The MOSFET sensitive element is a dual transistor with one element made of Silicon and the other SiO₂. Figure 7-2 plots the energy response of these two detectors along with the PSD relative to air to demonstrate their energy response correction factors at high and low keV energies. The TLD shares a similar energy response to the MOSFET dosimeter relative to air. Therefore, when both are calibrated free-in-air their response will be over-responsive
and overestimate the radiation absorbed dose. Although, the TLD seems to much less affected since the correction factors only varies less across the energy range of interest.

**Future Work**

**Future Work Project A**

This project focuses on constructing a PMT photon counter device that reduces the dead time, and more importantly, analyzes the counting losses that it the system is experiencing. The PMT overall is a fast responsive device that converts light into an electrical current. The conversion of charge into pulses is what ultimately limits the response time. The photon counter of the detector array studied in this work consists of an amplifier, comparator, pulse shaper, and counter. All of these devices are relatively inexpensive and with the circuit boards our department owns would allow our lab group to construct our own device; permitted we have an accessible PMT. The first part of this project would be to analyze the output of the PMT from different radiation sources to simulate the response of the PMT. These responses can be simulated from a function generator and connected to a circuit and viewed with an oscilloscope to decipher any issues with photon counting. Dr. James Schumacher, from the University of Florida, is a knowledgeable source and his lab has almost of all of the equipment that would be needed. The research is based entirely on constructing a circuit that optimizes the signal processing for photon detection in real time. Considerations should include: photon counting vs. signal integration, and time response of the electrical components.

**Future Work Project B**

This project is focused on rebuilding the dual scintillating detector using two inorganic scintillator phosphors, of which neither of demonstrate quenching effects.
These scintillators could be extracted from an x-ray film cassette or from a vendor. Ideally the use of low-Z scintillators would be used, which have K-edge below the common energies encountered in CT. However high k-edges have proven useful as well since most of the absorbed energy coincides with the effective energy of the x-ray and sensitivity to scatter which make their response less variable. The methods for characterizing such a detector are laid out in Chapter 3. In addition to this project, it would of interest to compare each detector’s free-in-air and in a scattering medium to quantify the over-response in reported doses by previous studies. It would also be useful to evaluate the energy dependence across a range of tube potentials (20 thru 200 kVp) to demonstrate the effects of quenching outside the 60 thru 140 kVp range.

**Future Work Project C**

Finally, there is now a physical validation methodology in place to verify Monte Carlo simulations on the computational pregnant female series utilizing TCM. At this moment Dr. Bolch’s research group has already created a TCM source term for a Toshiba Aquilion ONE scanner. One of the limits of the source model however, is that it can only perform retrospective dosimetry since the actual TCM algorithm is proprietary. Therefore images from the real CT scans are required to report the average/effective mAs per slice, which can be used to reconstruct the source profile and be used for the simulated dosimetry. Measurements following the methodology in this dissertation would be repeated on the scanner and given to Dr. Bolch’s research group to perform the Monte Carlo calculations. This would be a great project since both the data necessary for dosimetry, scanner reported CTDIvol values and DLP values, and physical measurements can be acquired by our research group that would vital
potentially leading to a general dosimetry model and validation of some of the results in this manuscript.
Figure 7-1. Plot of $\mu_{\text{en}}/\rho$ ratio between polystyrene and air in the low-energy range, taken from the National Institute of Standards and Technology (Ref 108). Linear fit was applied to the data points between 35 and 75 keV.
Figure 7-2. Plot of $\mu_{en}/\rho$ ratio between various dosimeters and air in the low-energy range, taken from the National Institute of Standards and Technology (Ref 108). LiF is the sensitive element of the TLD and silicon dioxide is that of the MOSFET dosimeter.
APPENDIX A
MATLAB CODE FOR GRAPHICAL INTERFACE

function varargout = CT_PMT(varargin)
% CT_PMT M-file for CT_PMT.fig
% CT_PMT, by itself, creates a new CT_PMT or raises the existing
% singleton*.
% H = CT_PMT returns the handle to a new CT_PMT or the handle to
% the existing singleton*.
% CT_PMT('CALLBACK',hObject,eventData,handles,...) calls the local
% function named CALLBACK in CT_PMT.M with the given input arguments.
% CT_PMT('Property','Value',...) creates a new CT_PMT or raises the
% existing singleton*. Starting from the left, property value pairs are
% applied to the GUI before CT_PMT_OpeningFcn gets called. An
% unrecognized property name or invalid value makes property application
% stop. All inputs are passed to CT_PMT_OpeningFcn via varargin.
% *See GUI Options on GUIDE's Tools menu. Choose "GUI allows only one
% instance to run (singleton)".
% See also: GUIDE, GUIDATA, GUIHANDLES

% Edit the above text to modify the response to help CT_PMT

% Last Modified by GUIDE v2.5 22-Jan-2014 00:09:29

% Begin initialization code - DO NOT EDIT
gui_Singleton = 1;
gui_State = struct('gui_Name',       mfilename, ...
                   'gui_Singleton',  gui_Singleton, ...
                   'gui_OpeningFcn', @CT_PMT_OpeningFcn, ...
                   'gui_OutputFcn',  @CT_PMT_OutputFcn, ...
                   'gui_LayoutFcn',  [] , ...
                   'gui_Callback',   []);
if nargin && ischar(varargin{1})
    gui_State.gui_Callback = str2func(varargin{1});
end
if nargout
    [varargout{1:nargout}] = gui_mainfcn(gui_State, varargin);
else
    gui_mainfcn(gui_State, varargin);
end
% End initialization code - DO NOT EDIT

% --- Executes just before CT_PMT is made visible.
% DEFAULT FUNCTION
function CT_PMT_OpeningFcn(hObject, eventdata, handles, varargin)
% This function has no output args, see OutputFcn.
% hObject    handle to figure
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)
% varargin   command line arguments to CT_PMT (see VARARGIN)

% Choose default command line output for CT_PMT
handles.output = hObject;

% Update handles structure
handles.data.count = 1;
set(hObject,'toolbar','figure'); %sets default toolbar onto GUI (top of figure)
guidata(hObject, handles);

% UIWAIT makes CT_PMT wait for user response (see UIRESUME)
% uiwait(handles.figure1);

% --- Outputs from this function are returned to the command line.
% DEFAULT FUNCTION
function varargout = CT_PMT_OutputFcn(hObject, eventdata, handles)
% varargout  cell array for returning output args (see VARARGOUT);
% hObject    handle to figure
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Get default command line output from handles structure
varargout{1} = handles.output;

%% Integration Time Input EDIT Text Boxes

function integration_time_input_PMT1_Callback(hObject, eventdata, handles)
% hObject    handle to integration_time_input_PMT1 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of integration_time_input_PMT1 as text
% str2double(get(hObject,'String')) returns contents of integration_time_input_PMT1 as a double
handles.data.int_time_PMT1=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
% edit text box for PMT 1 to input integration time
function integration_time_input_PMT1_CreateFcn(hObject, eventdata, handles)
% hObject    handle to integration_time_input_PMT1 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
% See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end

handles.data.int_time_PMT1=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on integration_time_input_PMT1 and none of its controls.
function integration_time_input_PMT1_KeyPressFcn(hObject, eventdata, handles)
    % hObject    handle to integration_time_input_PMT1 (see GCBO)
    % eventdata  structure with the following fields (see UICONTROL)
    %   Key: name of the key that was pressed, in lower case
    %   Character: character interpretation of the key(s) that was pressed
    %   Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
    % handles    structure with handles and user data (see GUIDATA)

    % Hints: get(hObject,'String') returns contents of
    % integration_time_input_PMT2 as text
    % str2double(get(hObject,'String')) returns contents of
    % integration_time_input_PMT2 as a double
    handles.data.int_time_PMT2=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
function integration_time_input_PMT2_CreateFcn(hObject, eventdata, handles)
    % hObject    handle to integration_time_input_PMT2 (see GCBO)
    % eventdata  reserved - to be defined in a future version of MATLAB
    % handles    empty - handles not created until after all CreateFcns called

    % Hint: edit controls usually have a white background on Windows.
    %       See ISPC and COMPUTER.
    if ispc && isequal(get(hObject,'BackgroundColor'),
get(0,'defaultUicontrolBackgroundColor'))
        set(hObject,'BackgroundColor','white');
    end

    handles.data.int_time_PMT2=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on integration_time_input_PMT2 and none of its controls.
function integration_time_input_PMT2_KeyPressFcn(hObject, eventdata, handles)
    % hObject    handle to integration_time_input_PMT2 (see GCBO)
    % eventdata  structure with the following fields (see UICONTROL)
    %   Key: name of the key that was pressed, in lower case
    %   Character: character interpretation of the key(s) that was pressed
% Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
% handles structure with handles and user data (see GUIDATA)

%%% Dead Time Edit Text Boxes

function dead_time_input_PMT1_Callback(hObject, eventdata, handles)
% hObject    handle to dead_time_input_PMT1 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of dead_time_input_PMT1 as text
%        str2double(get(hObject,'String')) returns contents of
% handles data.dead_time_PMT1=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
function dead_time_input_PMT1_CreateFcn(hObject, eventdata, handles)
% hObject    handle to dead_time_input_PMT1 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
% See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
handles.data.dead_time_PMT1=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on dead_time_input_PMT1 and none of its controls.
function dead_time_input_PMT1_KeyPressFcn(hObject, eventdata, handles)
% hObject    handle to dead_time_input_PMT1 (see GCBO)
% eventdata  structure with the following fields (see UICONTROL)
%   Key: name of the key that was pressed, in lower case
%   Character: character interpretation of the key(s) that was pressed
%   Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
% handles    structure with handles and user data (see GUIDATA)

function dead_time_input_PMT2_Callback(hObject, eventdata, handles)
% hObject    handle to dead_time_input_PMT2 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of dead_time_input_PMT2 as text
%        str2double(get(hObject,'String')) returns contents of
% handles data.dead_time_PMT2=str2double(get(hObject,'String'));
guidata(hObject, handles);
% --- Executes during object creation, after setting all properties.
function dead_time_input_PMT2_CreateFcn(hObject, eventdata, handles)
% hObject    handle to dead_time_input_PMT2 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
% See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
handles.data.dead_time_PMT2=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on dead_time_input_PMT2 and none of
% its controls.
function dead_time_input_PMT2_KeyPressFcn(hObject, eventdata, handles)
% hObject    handle to dead_time_input_PMT2 (see GCBO)
% eventdata  structure with the following fields (see UICONTROL)
%   Key: name of the key that was pressed, in lower case
%   Character: character interpretation of the key(s) that was pressed
%   Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
% handles    structure with handles and user data (see GUIDATA)

function dead_time_input_PMT3_Callback(hObject, eventdata, handles)
% hObject    handle to dead_time_input_PMT3 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of dead_time_input_PMT3 as
% text
% str2double(get(hObject,'String')) returns contents of
% dead_time_input_PMT3 as a double
handles.data.dead_time_PMT3=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
function dead_time_input_PMT3_CreateFcn(hObject, eventdata, handles)
% hObject    handle to dead_time_input_PMT3 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
% See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
handles.data.dead_time_PMT3=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on dead_time_input_PMT3 and none of
% its controls.
function dead_time_input_PMT3_KeyPressFcn(hObject, eventdata, handles)
function dead_time_input_PMT4_Callback(hObject, eventdata, handles)
    handles.data.dead_time_PMT4=str2double(get(hObject,'String'));
guidata(hObject, handles);

function dead_time_input_PMT4_CreateFcn(hObject, eventdata, handles)
    handles.data.dead_time_PMT4=str2double(get(hObject,'String'));
guidata(hObject, handles);

function dead_time_input_PMT4_KeyPressFcn(hObject, eventdata, handles)
    handles.data.dead_time_PMT4=str2double(get(hObject,'String'));
guidata(hObject, handles);

function dead_time_input_PMT5_Callback(hObject, eventdata, handles)
    handles.data.dead_time_PMT5=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
function dead_time_input_PMT5_CreateFcn(hObject, eventdata, handles)
% hObject    handle to dead_time_input_PMT5 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
%       See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
    get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
handles.data.dead_time_PMT5=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on dead_time_input_PMT5 and none of
% its controls.
function dead_time_input_PMT5_KeyPressFcn(hObject, eventdata, handles)
% hObject    handle to dead_time_input_PMT5 (see GCBO)
% eventdata  structure with the following fields (see UICONTROL)
%   Key: name of the key that was pressed, in lower case
%   Character: character interpretation of the key(s) that was pressed
%   Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
% handles    structure with handles and user data (see GUIDATA)

handles.data.dead_time_PMT5=str2double(get(hObject,'String'));
guidata(hObject, handles);

%%%%%%%%%%%%%%%% REF CALIBRATION FACTORS %%%%%%%%%%%%%%%%%

function ref_cal_fact_PMT1_Callback(hObject, eventdata, handles)
% hObject    handle to ref_cal_fact_PMT1 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of ref_cal_fact_PMT1 as text
% str2double(get(hObject,'String')) returns contents of ref_cal_fact_PMT1 as a double
handles.data.ref_cal_fact_PMT1=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
function ref_cal_fact_PMT1_CreateFcn(hObject, eventdata, handles)
% hObject    handle to ref_cal_fact_PMT1 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
%       See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
    get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
handles.data.ref_cal_fact_PMT1=str2double(get(hObject,'String'));
guidata(hObject, handles);
% --- Executes on key press with focus on ref_cal_fact_PMT1 and none of its controls.
function ref_cal_fact_PMT1_KeyPressFcn(hObject, eventdata, handles)
    hObject    handle to ref_cal_fact_PMT1 (see GCBO)
    eventdata  structure with the following fields (see UICONTROL)
    % Key: name of the key that was pressed, in lower case
    % Character: character interpretation of the key(s) that was pressed
    % Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
    % handles structure with handles and user data (see GUIDATA)

function ref_cal_fact_PMT2_Callback(hObject, eventdata, handles)
    hObject    handle to ref_cal_fact_PMT2 (see GCBO)
    eventdata  reserved - to be defined in a future version of MATLAB
    % handles structure with handles and user data (see GUIDATA)

    % Hints: get(hObject,'String') returns contents of ref_cal_fact_PMT2 as text
    % str2double(get(hObject,'String')) returns contents of ref_cal_fact_PMT2 as a double
    handles.data.ref_cal_fact_PMT2=str2double(get(hObject,'String'));
    guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
function ref_cal_fact_PMT2_CreateFcn(hObject, eventdata, handles)
    hObject    handle to ref_cal_fact_PMT2 (see GCBO)
    eventdata  reserved - to be defined in a future version of MATLAB
    % handles empty - handles not created until after all CreateFcns called

    % Hint: edit controls usually have a white background on Windows.
    %       See ISPC and COMPUTER.
    if ispc && isequal(get(hObject,'BackgroundColor'),
        get(0,'defaultUicontrolBackgroundColor'))
        set(hObject,'BackgroundColor','white');
    end
    handles.data.ref_cal_fact_PMT2=str2double(get(hObject,'String'));
    guidata(hObject, handles);

% --- Executes on key press with focus on ref_cal_fact_PMT2 and none of its controls.
function ref_cal_fact_PMT2_KeyPressFcn(hObject, eventdata, handles)
    hObject    handle to ref_cal_fact_PMT2 (see GCBO)
    eventdata  structure with the following fields (see UICONTROL)
    % Key: name of the key that was pressed, in lower case
    % Character: character interpretation of the key(s) that was pressed
    % Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
    % handles structure with handles and user data (see GUIDATA)

++++++++++++++++ PRIMARY BEAM CALIBRATION FACTORS +++++++++++++++++

function prim_cal_fact_PMT1_Callback(hObject, eventdata, handles)
    hObject    handle to prim_cal_fact_PMT1 (see GCBO)
    eventdata  reserved - to be defined in a future version of MATLAB
    % handles structure with handles and user data (see GUIDATA)
% Hints: get(hObject,'String') returns contents of prim_cal_fact_PMT1 as text
% str2double(get(hObject,'String')) returns contents of
% prim_cal_fact_PMT1 as a double
% --- Executes during object creation, after setting all properties.
handles.data.prim_cal_fact_PMT1=str2double(get(hObject,'String'));
guidata(hObject, handles);

function prim_cal_fact_PMT1_CreateFcn(hObject, eventdata, handles)
% hObject    handle to prim_cal_fact_PMT1 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
% See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
handles.data.prim_cal_fact_PMT1=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on prim_cal_fact_PMT1 and none of its controls.
function prim_cal_fact_PMT1_KeyPressFcn(hObject, eventdata, handles)
% hObject    handle to prim_cal_fact_PMT1 (see GCBO)
% eventdata  structure with the following fields (see UICONTROL)
%   Key: name of the key that was pressed, in lower case
%   Character: character interpretation of the key(s) that was pressed
%   Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
% handles    structure with handles and user data (see GUIDATA)

function prim_cal_fact_PMT2_Callback(hObject, eventdata, handles)
% hObject    handle to prim_cal_fact_PMT2 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of prim_cal_fact_PMT2 as text
% str2double(get(hObject,'String')) returns contents of
% prim_cal_fact_PMT2 as a double

handles.data.prim_cal_fact_PMT2=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
function prim_cal_fact_PMT2_CreateFcn(hObject, eventdata, handles)
% hObject    handle to prim_cal_fact_PMT2 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
% See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
get(0,'defaultUicontrolBackgroundColor'))

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set(hObject,'BackgroundColor','white');
end
handles.data.prim_cal_fact_PMT2=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on prim_cal_fact_PMT2 and none of its controls.
function prim_cal_fact_PMT2_KeyPressFcn(hObject, eventdata, handles)
% hObject    Handle to prim_cal_fact_PMT2 (see GCBO)
% eventdata  structure with the following fields (see UICONTROL)
%   Key: name of the key that was pressed, in lower case
%   Character: character interpretation of the key(s) that was pressed
%   Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
% handles    structure with handles and user data (see GUIDATA)

%%%%%%%%%%%%%%%%%%SECONDARY CALIBRATION FACTORS %%%%%%%%%%%%%%%%%

function sec_cal_fact_PMT1_Callback(hObject, eventdata, handles)
% hObject    handle to sec_cal_fact_PMT1 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of sec_cal_fact_PMT1 as text
%        str2double(get(hObject,'String')) returns contents of sec_cal_fact_PMT1 as a double
handles.data.sec_cal_fact_PMT1=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
function sec_cal_fact_PMT1_CreateFcn(hObject, eventdata, handles)
% hObject    Handle to sec_cal_fact_PMT1 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
%       See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
    get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
handles.data.sec_cal_fact_PMT1=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on sec_cal_fact_PMT1 and none of its controls.
function sec_cal_fact_PMT1_KeyPressFcn(hObject, eventdata, handles)
% hObject    handle to sec_cal_fact_PMT1 (see GCBO)
% eventdata  structure with the following fields (see UICONTROL)
%   Key: name of the key that was pressed, in lower case
%   Character: character interpretation of the key(s) that was pressed
%   Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
% handles    structure with handles and user data (see GUIDATA)
function sec_cal_fact_PMT2_Callback(hObject, eventdata, handles)
% hObject    handle to sec_cal_fact_PMT2 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of sec_cal_fact_PMT2 as text
% str2double(get(hObject,'String')) returns contents of sec_cal_fact_PMT2 as a double
handles.data.sec_cal_fact_PMT2=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
function sec_cal_fact_PMT2_CreateFcn(hObject, eventdata, handles)
% hObject    handle to sec_cal_fact_PMT2 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
% See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
    get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
handles.data.sec_cal_fact_PMT2=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on sec_cal_fact_PMT2 and none of its controls.
function sec_cal_fact_PMT2_KeyPressFcn(hObject, eventdata, handles)
% hObject    handle to sec_cal_fact_PMT2 (see GCBO)
% eventdata  structure with the following fields (see UICONTROL)
%   Key: name of the key that was pressed, in lower case
%   Character: character interpretation of the key(s) that was pressed
%   Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
% handles    structure with handles and user data (see GUIDATA)

%%%%%%%%%%%%%%%%%%%%%%%% COUNT RATE THRESHOLD %%%%%%%%%%%%%%%%%%%%%%%%%

function cr_thresh_PMT1_Callback(hObject, eventdata, handles)
% hObject    handle to cr_thresh_PMT1 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of cr_thresh_PMT1 as text
% str2double(get(hObject,'String')) returns contents of cr_thresh_PMT1 as a double
handles.data.cr_thresh_PMT1=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
function cr_thresh_PMT1_CreateFcn(hObject, eventdata, handles)
% hObject    handle to cr_thresh_PMT1 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
%       See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
                 get(0,'defaultUicontrolBackgroundColor'))
  set(hObject,'BackgroundColor','white');
end

handles.data.cr_thresh_PMT1=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on cr_thresh_PMT1 and none of its controls.
function cr_thresh_PMT1_KeyPressFcn(hObject, eventdata, handles)
  hObject    handle to cr_thresh_PMT1 (see GCBO)
  eventdata  structure with the following fields (see UICONTROL)
  Key: name of the key that was pressed, in lower case
  Character: character interpretation of the key(s) that was pressed
  Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
  handles    structure with handles and user data (see GUIDATA)

function crt_thresh_PMT2_Callback(hObject, eventdata, handles)
  hObject    handle to crt_thresh_PMT2 (see GCBO)
  eventdata  structure with the following fields (see UICONTROL)
  Key: name of the key that was pressed, in lower case
  Character: character interpretation of the key(s) that was pressed
  Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
  handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of crt_thresh_PMT2 as text
%        str2double(get(hObject,'String')) returns contents of
crt_thresh_PMT2 as a double
handles.data.crt_thresh_PMT2=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
function crt_thresh_PMT2_CreateFcn(hObject, eventdata, handles)
  hObject    handle to crt_thresh_PMT2 (see GCBO)
  eventdata  reserved - to be defined in a future version of MATLAB
  handles    structure with handles and user data (see GUIDATA)

% Hint: edit controls usually have a white background on Windows.
%       See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
                   get(0,'defaultUicontrolBackgroundColor'))
  set(hObject,'BackgroundColor','white');
end
handles.data.crt_thresh_PMT2=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on crt_thresh_PMT2 and none of its controls.
function crt_thresh_PMT2_KeyPressFcn(hObject, eventdata, handles)
  hObject    handle to crt_thresh_PMT2 (see GCBO)
  eventdata  structure with the following fields (see UICONTROL)
  Key: name of the key that was pressed, in lower case
  Character: character interpretation of the key(s) that was pressed
% Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
% handles    structure with handles and user data (see GUIDATA)

% DUAL SCINT CONTROLS

function center_ratio_Callback(hObject, eventdata, handles)
    hObject    handle to center_ratio (see GCBO)
    eventdata  reserved - to be defined in a future version of MATLAB
    handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of center_ratio as text
%        str2double(get(hObject,'String')) returns contents of center_ratio as a double
handles.data.center_ratio=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
function center_ratio_CreateFcn(hObject, eventdata, handles)
    hObject    handle to center_ratio (see GCBO)
    eventdata  reserved - to be defined in a future version of MATLAB
    handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
%       See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
                   get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
handles.data.center_ratio=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on center_ratio and none of its controls.
function center_ratio_KeyPressFcn(hObject, eventdata, handles)
    hObject    handle to center_ratio (see GCBO)
    eventdata  structure with the following fields (see UITC
    % Key: name of the key that was pressed, in lower case
    % Character: character interpretation of the key(s) that was pressed
    % Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
    % handles    structure with handles and user data (see GUIDATA)

function ratio_cal_fact_Callback(hObject, eventdata, handles)
    hObject    handle to ratio_cal_fact (see GCBO)
    eventdata  reserved - to be defined in a future version of MATLAB
    handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of ratio_cal_fact as text
%        str2double(get(hObject,'String')) returns contents of ratio_cal_fact as a double
handles.data.ratio_cal_fact=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes during object creation, after setting all properties.
function ratio_cal_fact_CreateFcn(hObject, eventdata, handles)
% hObject    handle to ratio_cal_fact (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
% See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
    get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
handles.data.ratio_cal_fact=str2double(get(hObject,'String'));
guidata(hObject, handles);

% --- Executes on key press with focus on ratio_cal_fact and none of its controls.
function ratio_cal_fact_KeyPressFcn(hObject, eventdata, handles)
% hObject    handle to ratio_cal_fact (see GCBO)
% eventdata  structure with the following fields (see UICONTROL)
%   Key: name of the key that was pressed, in lower case
%   Character: character interpretation of the key(s) that was pressed
%   Modifier: name(s) of the modifier key(s) (i.e., control, shift) pressed
% handles    structure with handles and user data (see GUIDATA)

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
% PMT/USER Control Buttons

% --- Executes on button press in Start.
function Start_Callback(hObject, eventdata, handles)
% hObject    handle to Start (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Set Start to value of 1 so as long as activated data will be aquired
set(handles.Start,'UserData',1);
%integration time in binary format to send to each available PMT
inttime_1=uint8([128,0,0,0,handles.data.int_time_PMT1/10]);
inttime_2=uint8([128,0,0,0,handles.data.int_time_PMT2/10]);

if handles.data.count == 1 % open up conmunication with PMT's 1 thru 5
    handles.data.count = 0;
    measurenum = uint8([129,0,0,0,1]); % measurement number to send to the
    PMT (usually 1)
    handles.data.startcmd = uint8([132,0,0,0,1]); % command to begin counting
    sent to PMT
    handles.data.stopcmd = uint8([132,0,0,0,0]); % command to stop counting
    sent to PMT
PMT 1
% open serial port to PMT 1
handles.data.serialobj1=serial('COM101','BaudRate', 9600, 'Parity',
'none','InputBufferSize', 30000);
fclose(handles.data.serialobj1);
fwrite(handles.data.serialobj1,handles.data.stopcmd);
fwrite(handles.data.serialobj1,measurenum);
fwrite(handles.data.serialobj1,inttime_1);
fwrite(handles.data.serialobj1,handles.data.startcmd);

PMT 2
% open serial port to PMT 2
handles.data.serialobj2=serial('COM102','BaudRate', 9600, 'Parity',
'none','InputBufferSize', 30000);
fclose(handles.data.serialobj2);
fwrite(handles.data.serialobj2,handles.data.stopcmd);
fwrite(handles.data.serialobj2,measurenum);
fwrite(handles.data.serialobj2,inttime_1);
fwrite(handles.data.serialobj2,handles.data.startcmd);

PMT 3
% open serial port to PMT 3
handles.data.serialobj3=serial('COM103','BaudRate', 9600, 'Parity',
'none','InputBufferSize', 30000);
fclose(handles.data.serialobj3);
fwrite(handles.data.serialobj3,handles.data.stopcmd);
fwrite(handles.data.serialobj3,measurenum);
fwrite(handles.data.serialobj3,inttime_1);
fwrite(handles.data.serialobj3,handles.data.startcmd);

PMT 4
% open serial port to PMT 4
handles.data.serialobj4=serial('COM104','BaudRate', 9600, 'Parity',
'none','InputBufferSize', 30000);
fclose(handles.data.serialobj4);
fwrite(handles.data.serialobj4,handles.data.stopcmd);
fwrite(handles.data.serialobj4,measurenum);
fwrite(handles.data.serialobj4,inttime_2);
fwrite(handles.data.serialobj4,handles.data.startcmd);

PMT 5
% open serial port to PMT 5
handles.data.serialobj5=serial('COM105','BaudRate', 9600, 'Parity',
'none','InputBufferSize', 30000);
fclose(handles.data.serialobj5);
fwrite(handles.data.serialobj5, handles.data.stopcmd);
fwrite(handles.data.serialobj5,measurenum);
fwrite(handles.data.serialobj5,inttime_2);
fwrite(handles.data.serialobj5,handles.data.startcmd);

else   % used to flush data and resend int time to get back more data

% PMT 1

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flushinput(handles.data.serialobj1);
fwrite(handles.data.serialobj1,inttime_1);
fwrite(handles.data.serialobj1,handles.data.startcmd);

% PMT 2
flushinput(handles.data.serialobj2);
fwrite(handles.data.serialobj2,inttime_1);
fwrite(handles.data.serialobj2,handles.data.startcmd);

% PMT 3
flushinput(handles.data.serialobj3);
fwrite(handles.data.serialobj3,inttime_1);
fwrite(handles.data.serialobj3,handles.data.startcmd);

% PMT 4
flushinput(handles.data.serialobj4);
fwrite(handles.data.serialobj4,inttime_2);
fwrite(handles.data.serialobj4,handles.data.startcmd);

% PMT 5
flushinput(handles.data.serialobj5);
fwrite(handles.data.serialobj5,inttime_2);
fwrite(handles.data.serialobj5,handles.data.startcmd);
end

A=[];
B=[];
C=[];
D=[];
E=[];
handles.data.output1 = [];
handles.data.output2 = [];

handles.data.out1= [];
handles.data.out2= [];
handles.data.out3= [];
handles.data.out4= [];
handles.data.out5= [];

handles.data.x1 = [handles.data.int_time_PMT1/1000];
handles.data.x2 = [handles.data.int_time_PMT2/1000];

i=1;
j=1;
k=1;
m=1;
n=1;

total1=0;
total2=0;

% set PMT initial counts (total and integral) to 0
set(handles.PMT1_counts, 'string', num2str(0));
set(handles.PMT1_total, 'string', num2str(0));
set(handles.PMT2_counts, 'string', num2str(0));
set(handles.PMT2_total, 'string', num2str(0));

guidata(hObject, handles);

while (get(handles.Start, 'UserData') == 1)
    if handles.data.serialobj3.BytesAvailable >= 5
        C=fread(handles.data.serialobj3, 5);
        if C(1) == 65
            handles.data.out3(k) = C(2)*256^3 + C(3)*256^2 + C(4)*256 + C(5);
            handles.data.output1(3,k) = ((handles.data.out3(k)/(handles.data.int_time_PMT1/1000))/(1-
            handles.data.dead_time_PMT3*(handles.data.out3(k)/(handles.data.int_time_PMT1/
            1000))))*(handles.data.int_time_PMT1/1000);
        elseif C(1) == 66
            set(handles.PMT1_counts, 'string', 'Overflow');
            drawnow
            Break
        else
            set(handles.PMT1_counts, 'string', 'Error!');
            drawnow
            Break
        end
        k=k+1;
    end
    drawnow
end

%Gad Fiber

if handles.data.serialobj2.BytesAvailable >= 5
    B=fread(handles.data.serialobj2, 5);
    if B(1) == 65
        handles.data.out2(j) = B(2)*256^3 + B(3)*256^2 + B(4)*256 + B(5);
        handles.data.output1(2,j) = ((handles.data.out2(j)/(handles.data.int_time_PMT1/1000))/(1-
        handles.data.dead_time_PMT2*(handles.data.out2(j)/(handles.data.int_time_PMT1/
        1000))))*(handles.data.int_time_PMT1/1000);
        handles.data.output1(5,j)=(handles.data.output1(2,j)-
        handles.data.output1(3,j))/(handles.data.int_time_PMT1/1000);
    elseif B(1) == 66
        set(handles.PMT1_counts, 'string', 'Overflow');
        drawnow
        Break
    else
        set(handles.PMT1_counts, 'string', 'Error!');
        drawnow
        Break
    end
    j=j+1;
    drawnow

end

% Plastic Fiber
if handles.data.serialobj1.BytesAvailable >= 5
    A=fread(handles.data.serialobj1, 5);
    if A(1) == 65
        handles.data.upper=0.05(handles.data.center_ratio+handles.data.center_ratio;
        handles.data.lower=handles.data.center_ratio-
        0.05*handles.data.center_ratio;
        handles.data.outl(i) = A(2)*256^3+A(3)*256^2+A(4)*256+A(5);
        handles.data.output1(l,i) =
        ((handles.data.outl(i)/(handles.data.int_time_PMT1/1000)))/(1-
        handles.data.dead_time_PMT1*(hands.data.data.outl(i)/(handles.data.int_time_PMT1
        /1000))))*(hands.data.data.int_time_PMT1/1000);
        handles.data.output1(4,i)=(hands.data.data.output1(1,i)+
        handles.data.data.ref_cal_fact_PMT1*hands.data.data.output1(3,i))/(hands.data.data.int_t
        ime_PMT1/1000);
        handles.data.output1(6,i)=hands.data.data.output1(4,i)/hands.data.data.output1(5,i)
        ;
        if handles.data.output1(1,i) > 15 ; handles.data.output1(4,i) >
        handles.data.data.cr_thresh_PMT1 ; handles.data.data.ouput1(6,i) < handles.data.data.upper ;
        handles.data.data.ouput1(6,i) > handles.data.data.lower ;
        total1=total1+(hands.data.data.int_time_PMT1/1000)*hands.data.data.ouput1(4,i)*hand
        les.data.data.prim_cal_fact_PMT1;
    end

    %%%%%%%%% OUTSIDE RATIO
    if handles.data.data.ouput1(1,i) > 15 ; handles.data.data.ouput1(4,i) >
    handles.data.data.cr_thresh_PMT1 ; handles.data.data.ouput1(6,i) < handles.data.data.lower ;
    total1=total1+(hands.data.data.int_time_PMT1/1000)*hands.data.data.ouput1(4,i)*hand
    les.data.data.prim_cal_fact_PMT1*hands.data.data.ratio_cal_fact*hands.data.data.output(6
    ,i);
    end

    if handles.data.data.ouput1(1,i) > 15 ; handles.data.data.ouput1(4,i) >
    handles.data.data.cr_thresh_PMT1 ; handles.data.data.ouput1(6,i) > handles.data.data.upper ;
    total1=total1+(hands.data.data.int_time_PMT1/1000)*hands.data.data.ouput1(4,i)*hand
    les.data.data.prim_cal_fact_PMT1*hands.data.data.ratio_cal_fact*hands.data.data.output(6
    ,i);
    end

    %%%%%%%%%%%%%% LOW COUNT RATE
    if handles.data.data.ouput1(1,i) > 15 ; handles.data.data.ouput1(4,i) <
    handles.data.data.cr_thresh_PMT1

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total1=total1+(handles.data.int_time_PMT1/1000)*handles.data.output1(4,i)*handles.data.sec_cal_fact_PMT1;

end

set(handles.PMT1_counts, 'string',
num2str(handles.data.output1(6,i)));
if i > 1
    handles.data.x1(i) = handles.data.x1(i-1)+handles.data.int_time_PMT1/1000;
end
plot(handles.axes1,handles.data.x1,handles.data.output1(6,:),
'b');
xlabel('Time (s)');
ylabel('Ratio');
elseif A(1) == 65
    handles.data.out5(n) = E(2)*256^3+E(3)*256^2+E(4)*256+E(5);
    handles.data.output2(2,n) = ((handles.data.out5(n)/(handles.data.int_time_PMT2/1000)))/(1-
(handles.data.dead_time_PMT5*(handles.data.out5(n)/(handles.data.int_time_PMT2
/1000))));
elseif E(1) == 66
    set(handles.PMT1_counts,'string','Overflow');
drawnow
    Break
else
    set(handles.PMT1_counts, 'string', 'Error!');
drawnow
    Break
end
i=i+1;
set(handles.PMT1_total, 'string', num2str(total1));
drawnow
end

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%
if handles.data.serialobj5.BytesAvailable >= 5
    E=fread(handles.data.serialobj5, 5);
    if E(1) == 65
        handles.data.out5(n) = E(2)*256^3+E(3)*256^2+E(4)*256+E(5);
        handles.data.output2(2,n) = ((handles.data.out5(n)/(handles.data.int_time_PMT2/1000)))/(1-
(handles.data.dead_time_PMT5*(handles.data.out5(n)/(handles.data.int_time_PMT2
/1000))));
    elseif E(1) == 66
        set(handles.PMT2_counts,'string','Overflow');
drawnow
        Break
    else
        set(handles.PMT2_counts, 'string', 'Error!');
drawnow
        Break
    end
    n=n+1;
end

drawnow
%% Linear FOC fiber
    if handles.data.serialobj4.BytesAvailable >= 5
        D=fread(handles.data.serialobj4, 5);
        if D(1) == 65
            handles.data.out4(m) = D(2)*256^3+D(3)*256^2+D(4)*256+D(5);
            handles.data.output2(1,m) = ((handles.data.out4(m)/(handles.data.int_time_PMT2/1000)))/(1-
                handles.data.dead_time_PMT4*(handles.data.out4(m)/(handles.data.int_time_PMT2
                /1000))))*(handles.data.int_time_PMT2/1000);
            handles.data.output2(3,m)=(handles.data.output2(1,m)-
                handles.data.ref_cal_fact_PMT2*handles.data.output2(2,m))/(handles.data.int_t
                ime_PMT2/1000);

            %%%%%%%%% DIRECT BEAM
            if handles.data.output2(1,m) > 15 ; handles.data.output2(3,m) >
                handles.data.crt_thresh_PMT2 ;

                total2=total2+handles.data.output2(3,m)*(handles.data.int_time_PMT2/1000)*hand
                les.data.prim_cal_fact_PMT2;
            end

            %%%%%%%%% SCATTER BEAM
            if handles.data.output2(1,m) > 15 ; handles.data.output2(3,m) <
                handles.data.crt_thresh_PMT2 ;

                total2=total2+handles.data.output2(3,m)*(handles.data.int_time_PMT2/1000)*hand
                les.data.sec_cal_fact_PMT2;
            end

            set(handles.PMT2_counts, 'string',
                num2str(handles.data.output2(3,m)));
            if m > 1
                handles.data.x2(m) = handles.data.x2(m-
                1)+handles.data.int_time_PMT2/1000;
            end
            plot(handles.axes2,handles.data.x2,handles.data.output2(3,:),
                'b');
            xlabel('Time (s)');
            ylabel('Count Rate');
            elseif D(1) == 66
                set(handles.PMT2_counts,'string','Overflow');
                drawnow
                Break
            else
                set(handles.PMT2_counts, 'string', 'Error!');
                drawnow
                Break
            end
            m=m+1;
            set(handles.PMT2_total, 'string', num2str(total2));
            drawnow
        end
    end
end

guida(hObject, handles);
% End of Start Button

% Stop Button
% --- Executes on button press in Stop.
function Stop_Callback(hObject, eventdata, handles)
% hObject    handle to Stop (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)
set(handles.Start,'UserData',0);
fwrite(handles.data.serialobj1,handles.data.stopcmd);
fwrite(handles.data.serialobj2,handles.data.stopcmd);
fwrite(handles.data.serialobj3,handles.data.stopcmd);
fwrite(handles.data.serialobj4,handles.data.stopcmd);
fwrite(handles.data.serialobj5,handles.data.stopcmd);

guida(hObject, handles);

% Disconnect Button
% --- Executes on button press in Disconnect.
function Disconnect_Callback(hObject, eventdata, handles)
% hObject    handle to Disconnect (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)
handles.data.output1 = [0];
handles.data.x1 = [0];

handles.data.output2 = 0;
handles.data.x2 = 0;

% PMT 1
fclose(handles.data.serialobj1);
axes(handles.axes1);
plot(handles.data.x1,handles.data.output1);
axis([0 1 0 1])
drawnow

% PMT 2
fclose(handles.data.serialobj2);
drawnow

% PMT 3
fclose(handles.data.serialobj3);
drawnow

% PMT 4
fclose(handles.data.serialobj4);
axes(handles.axes2);
plot(handles.data.x2,handles.data.output2);
axis([0 1 0 1])

drawnow

% PMT 5
fclose(handles.data.serialobj5);
drawnow

handles.data.count = 1;
guida(hObject, handles);

% File name
function Filename_Callback(hObject, eventdata, handles)
    % hObject    handle to Filename (see GCBO)
    % eventdata  reserved - to be defined in a future version of MATLAB
    % handles    structure with handles and user data (see GUIDATA)

    % Hints: get(hObject,'String') returns contents of Filename as text
    %       str2double(get(hObject,'String')) returns contents of Filename as a
double
    handles.data.Filename= get(hObject,'String');
guida(hObject, handles);

% --- Executes during object creation, after setting all properties.
function Filename_CreateFcn(hObject, eventdata, handles)
    % hObject    handle to Filename (see GCBO)
    % eventdata  reserved - to be defined in a future version of MATLAB
    % handles    empty - handles not created until after all CreateFcns called

    % Hint: edit controls usually have a white background on Windows.
    %       See ISPC and COMPUTER.
    if ispc && isequal(get(hObject,'BackgroundColor'),
        get(0,'defaultUicontrolBackgroundColor'))
        set(hObject,'BackgroundColor','white');
    end
    handles.data.Filename= get(hObject,'String');
guida(hObject, handles);

%Save to text file button Button

% --- Executes on button press in Filewrite.
function Filewrite_Callback(hObject, eventdata, handles)
    % hObject    handle to Filewrite (see GCBO)
    % eventdata  reserved - to be defined in a future version of MATLAB
    % handles    structure with handles and user data (see GUIDATA)

    a=length(handles.data.output1);
b=length(handles.data.output2);

    M=zeros(11,a);
    M(1,:)=handles.data.x1;
    M(2:7,:)=handles.data.output1;
M(8,:)=handles.data.x2;
M(9:11,:)=handles.data.output2;

text_file=fopen(handles.data.Filename,'wt');
fprintf(text_file,'%s	 %s	 %s	 %s	 %s	 %s	 %s	 %s	 %s	 %s	
','T(s)','PMT1','PMT2','PMT3','PSCR','GSCR','RATO','T(s)','PMT4','PMT5','LDCR');
fprintf(text_file,'%g	 %g	 %g	 %g	 %g	 %g	 %g	 %g	 %g	 %g	
',M);
guidata(hObject, handles);

function PMT1_counts_Callback(hObject, eventdata, handles)
% hObject    handle to PMT1_counts (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)
% Hints: get(hObject,'String') returns contents of PMT1_counts as text
%        str2double(get(hObject,'String')) returns contents of PMT1_counts as a double

figure;

function PMT1_counts_CreateFcn(hObject, eventdata, handles)
% hObject    handle to PMT1_counts (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

function PMT1_total_Callback(hObject, eventdata, handles)
% hObject    handle to PMT1_total (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)
% Hints: get(hObject,'String') returns contents of PMT1_total as text
%        str2double(get(hObject,'String')) returns contents of PMT1_total as a double

figure;

function PMT1_total_CreateFcn(hObject, eventdata, handles)
% hObject    handle to PMT1_total (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called
% Hint: edit controls usually have a white background on Windows.
% See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end

function PMT2_counts_Callback(hObject, eventdata, handles)
% hObject    handle to PMT2_counts (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of PMT2_counts as text
%        str2double(get(hObject,'String')) returns contents of PMT2_counts as a double

% --- Executes during object creation, after setting all properties.
function PMT2_counts_CreateFcn(hObject, eventdata, handles)
% hObject    handle to PMT2_counts (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
% See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end

function PMT2_total_Callback(hObject, eventdata, handles)
% hObject    handle to PMT2_total (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of PMT2_total as text
%        str2double(get(hObject,'String')) returns contents of PMT2_total as a double

% --- Executes during object creation, after setting all properties.
function PMT2_total_CreateFcn(hObject, eventdata, handles)
% hObject    handle to PMT2_total (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
% See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
The elongated scintillating fiber, when properly calibrated, measures the absorbed dose along its axis. By pairing two of these detectors orthogonal to one another, the dose in the imaging plane is more accurately measured, due to the nature of obtaining a one dimensional integral dose, combined with the additional information provided by the second detector. This type of dosimetry setup, as seen in Figure B-1, is very compatible with the UF anthropomorphic phantoms. Dosimeters are placed perpendicular to each other and intersect at the center of the organ. In this setup, the absorbed dose to the liver is being measured in one of the slices.

Figure B-1. Dosimeter alignment for performing organ dosimetry.
Organs lying in the axial plane can be modeled as 2-D geometric shapes such as a circle or ellipse, as demonstrated by early stylized phantoms. Over time, phantoms have evolved, and their organs have been rendered into complex shapes. The computational reference phantoms at the University of Florida (UF) are constructed using voxelized CT cross sectional images interpolated together and modified to fit the reference criteria. The UF anthropomorphic phantoms are derived from these hybrid phantoms and therefore contain structures that cannot be defined with a simple mathematical formula. However, it would not be unreasonable to treat these complex shapes as an ellipse or any other simple geometric shape defined by one or two parameters such as illustrated in Figure B-2. The ellipse shown below is defined by its radi $a$ and $b$.

\[ \text{Area} = \pi ab \]

Figure B-2. Geometry of an ellipse.

An ellipse is also a good geometric approximation for an organ because of its smooth exterior contours that resemble the surfaces of structures in the human body. This is illustrated in Figure B-1 which has an equivalent organ (yellow lines) designated for each of the major organs. Another reason why an ellipse is a good mathematical model is because the ellipse is defined by two parameters. If a pair of detectors with sensitive lengths $2a$ and $2b$ were aligned along minor and major axis of an organ, the average 2-D dose could be estimated using the measurements from each detector and
applying a correctional factor to each measurement. The average organ dose is represented as a function of the two dosimeter readings, which are calibrated to the average dose along their sensitive length corresponding to the dimensions of the organ (a and b).

\[
\overline{D}_{\text{org}} = f(D_a, D_b, a, b) \tag{B-1}
\]

Looking at Figure B-3, the dosimeters are placed on the major and minor axis of the ellipse. The dosimeters measure the average dose along their respective axis. By definition the best estimate of the average organ dose will involve a component of each measured dose. The weighting of each component is based on the length of the major and minor axis of the organ. Therefore we can write the average organ dose as:

\[
\overline{D}_{\text{org}} = f(a, b) \ast D_x + g(a, b) \ast D_y \tag{B-2}
\]

Where f and g represent weighting factors that account for the contribution of each axis to the overall area of the ellipse. For example an ellipse similar to the shape of a circle should equally weight the major and minor axis component of measured absorbed dose. An ellipse shaped like a submarine should favor the dose measured along its major axis.
Now, assume that each detector can capture the dose profile at a distance $dr$ from its axis as illustrated above. For an ellipsoid characterized along the x, y, and z axis by $a$, $b$, and $c$ as shown below where $a > b > c$ modeled as an organ where dosimeters 2a and 2b are used to capture organ dose data, the fractional dose capture by each detector is:

$$D_{\text{org}} = \frac{2\pi (dr)^2}{4\pi abc} a D_a + \frac{2\pi (dr)^2}{4\pi abc} b D_b$$  \hspace{1cm} (B-3)$$

However, since we want the sum of $f$ and $g$ to equal 1 we will divide by the common terms and after normalizing we get our final expression for the estimated absorbed dose:

$$\bar{D}_{\text{org}} = \frac{a}{a + b} D_a + \frac{b}{a + b} D_b$$  \hspace{1cm} (B-4)$$

Figure B-3. Dosimeters aligning to geometry of an ellipse.
The above equation is only valid for an organ with a uniform density. Equation B-4 was used to weight the response of each dosimeter to obtain an average organ/fetal dose.
LIST OF REFERENCES


Hamamatsu, Hamamatsu Photonics, 2007.


91. E.G. Yukihara, S.W. McKeever, Optically stimulated luminescence: fundamentals and applications. (John Wiley & Sons Ltd, United Kingdom, 2011).


BIOGRAPHICAL SKETCH

Matthew Robert Hoerner was born in Pittsburgh, Pennsylvania, to Karen Flory and Richard Hoerner. He is one of two children, along with his younger brother, Michael. He graduated from Unionville High School in Kennett Square, Pennsylvania in 2005. Four years later he graduated from The Pennsylvania State University with a Bachelor of Science degree in nuclear engineering. In 2009, Matthew enrolled into the medical physics graduate program at the University of Florida where he completed his Master of Science and Doctor of Philosophy. After graduation, Matthew will begin his two year imaging physics residency at UF Health Shands Hospital in Gainesville, FL.