CHARACTERIZATION OF QUANTUM DOT SCINTILLATORS FOR MEDICAL DOSIMETRY

By

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To my Mom and Dad, who have never stopped believing in me, even when I doubted myself. Thanks to their faith in me, I found the strength to push through academic probation during my undergraduate years and now find myself flourishing and with academic honors in graduate school.
ACKNOWLEDGMENTS

I want to thank Dr. Hintenlang for his guidance and patience throughout this project, he was always willing to help but never gave an easy way out. Thanks to his approach, I have learned to become an independent learner.

I also thank Matthew Hoerner and Katherine Mittauer for the time they gave to help me learn to use laboratory equipment and the intricacies of constructing FOC dosimeters.
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CHARACTERIZATION OF QUANTUM DOT SCINTILLATORS FOR MEDICAL DOSIMETRY

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A near water equivalent scintillating solution of CdTe quantum dots with absorption peak at 567nm and fluorescence peak at 599nm was used to construct FOC dosimeters. The dosimeter was characterized for energy dependence, dose linearity, dose rate dependence, and reproducibility at the following energy ranges: 20-35kVp, 50-100kVp, and 4-18MV. The proposed use of this new dosimeter is for real-time in vivo dose monitoring in diagnostic imaging procedures. Several detectors were constructed and all had slightly different responses to each measurement. However, it was shown that a well-built detector can have well behaved energy dependence, linearity, and reproducibility. The best reproducibility achieved among the dosimeters gave a coefficient of variation=0.0227.
In vivo dosimetry (IVD) is the practice in which radiation dose is measured during medical procedures, most commonly through the use of TLD (Thermoluminescent dosimeters), diode, or MOSFET (Metal-oxide semiconductor field-effect transistors) detectors taped onto the surface of patients. Although in vivo dosimetry has been used in radiation therapy for many years now, it is currently not routine practice in diagnostic procedures despite recommendations from the ACR.\(^1\)

While the effects of high radiation doses have been thoroughly studied among the populations exposed at Hiroshima, Nagasaki, and Chernobyl, the effects of low doses of radiation have not been studied rigorously enough due to a lack of data. It is expected that the increase in CT and nuclear medicine exams over the last 25 years will result in an increase of radiation induced cancer. From 1980 to 2005 the number of annual CT exams in the United States rose from approximately 3,000,000 to 60,000,000 and the number of nuclear medicine exams rose from approximately 7,000,000 to 20,000,000.\(^2\) In an effort to monitor patient doses and to start building a data set for the effects of low levels of radiation, the National Institute of Health (NIH) is working with imaging equipment vendors to incorporate dose reports into the electronic medical record (EMR). These efforts will allow patients to keep track of their imaging history and their accumulated radiation doses as well as provide the necessary data to study the stochastic and deterministic effects of low radiation doses characteristic of diagnostic imaging procedures.\(^1\)
The importance of building a data set to study the epidemiological effects of low doses of radiation cannot be understated, but it is unclear whether or not IVD benefits patients on an individual basis. Even in therapeutic procedures, IVD is not always used. For instance, it may be used only on the first fraction to make sure a series of fractions do not repeat the same error. AAPM task group 42 recommends that institutions have an IVD system to identify any deviations from the prescription, but AAPM task group 60 emphasizes that an IVD system should be supplementary to a good QA program even though misadministrations are extremely rare at about 0.002% of all radiation therapy treatments.³

The truth about in vivo dosimetry is that it may not be cost-effective due to the low probability of misadministrations, but it has the potential to save lives. As is too often the case, mistakes can happen several times before they are caught, but with more frequent in vivo measurements the mistakes can be found sooner. In Panama, for example, 56 patients were mistreated from 2000-2001 resulting in 23 deaths.⁴

In the following subsections are descriptions of some of the more common in vivo dosimetry systems used in practice today.

**TLDs**

Thermoluminescent dosimeters are a type of radiation detector known as integrating passive detectors because they integrate the dose absorbed over the exposure time. TLDs work by trapping excited electrons in trapping centers and luminescence centers. Depending on the stability of these electron traps, the release of photons can be delayed longer. Materials with traps further from the bandgap edges are preferred because more energy can be stored for longer periods of time.⁵ After a TLD is exposed, the dose is measured by heating them until the trapped electrons are
stimulated to return to the valence band. Through the process of heated “de-excitation” the electrons release a photon in what is called thermoluminescence.

The most common materials used in clinical TLDs are LiF doped with Mg and Ti ions or LiF doped with Mg, Cu, and P ions. LiF is a favorable material for TLDs because depending on the ions used for doping, the efficiency at different dose levels changes. LiF:Mg,Ti is more efficient at high dose levels (>1Gy) and LiF:Mg,Cu,P is more efficient at dose levels below 1Gy.  

Some benefits of using TLDs are that they are independent of geometry or orientation and they can be made into very small chips, which gives them very good spatial resolution. The one major downfall of TLDs is that they are dose integrators and therefore cannot give real time dose measurements.

**Diodes**

Along with TLDs, diodes are the most commonly used detectors in IVD systems. Most diodes today are made of silicon doped with Phosphorus, Arsenic, Gallium, or Boron. Doping is used to increase the conductivity by increasing the number of electrons or holes in the valence band of the semiconductor. This makes the diode more sensitive to radiation because more charge can be collected per energy deposited. In fact, silicon diodes are 18,000 times more sensitive than air filled ion chambers because of their very high radiation current density. This, along with the fact that they are small, rugged, and read out dose in real time make diodes a very good option for an IVD system. Their downsides include that they can be damaged by energies higher than 2MeV and they are dependent on dose rate, temperature, field size, and beam energy.
MOSFETs

MOSFETs are another type of dose integrating dosimeter that measure dose by collecting charge created by ionizing radiation, the dose is linearly related to the shift in a threshold gate voltage.\(^5\)

MOSFETs have several weaknesses; one is that they have relatively low sensitivity, which decreases over the detector’s lifespan with increasing radiation damage. Another weakness is that their lifespan is very short; approximate lifespans for a typical MOSFET are around 100 Gy for 75 kVp x-rays and 300 Gy at 6 MV. Finally, MOSFETS have to be corrected for direction, dose rate, and energy dependence. Unfortunately, the only real benefit of MOSFETs is that they are small and therefore have good spatial resolution.\(^7\)

FOC dosimeters

Fiber optic coupled dosimeters are the least common used detectors in IVD systems and include the detector characterized in this study. FOCDs can be constructed with a variety of different materials, but the principle remains the same throughout. FOCDs work by coupling a scintillating material to a fiber optic cable that guides the scintillation light to a photomultiplier tube where the amount of light is measured. The intensity of the measured light is related to the absorbed dose in the scintillating material.

One advantage FOCDs have over other in vivo detectors is that many scintillating materials have near tissue equivalent properties that allow detection while minimizing perturbation of the beam. Like diodes, FOCDs also give real time dosimetry measurements. The only potential problems with FOCDs are that they are energy dependent (all of the above detectors are also energy dependent) and Cerenkov
radiation is produced in the cables producing a lot of noise. However, this can be corrected for by using chromatic filtering or background subtraction.

**Quantum Dots**

Quantum dots, also known as semiconductor nanocrystals, can be synthesized in many ways. Epitaxial growth modes such as Van der Merve, Volmer-Weber, and Stranski-Krastanow are layer by layer methods of synthesizing quantum dots on a crystal substrate. Newer methods used to produce colloidal quantum dots give better control of the size and shape formed during nucleation. Core-shell quantum dots are nanoparticles where one material is buried inside another, they are formed into nanoinclusions of about 500-1000 atoms. The most basic characteristic shared by all quantum dots is that they quantumly confine electrons in three dimensions, thus making it behave like an artificial atom.

The phenomenon of 3-dimensional quantum confinement leads to heavy dependence of band gap energy on the size of the quantum dot. This allows the electronic and optical properties of quantum dots to be highly tunable simply by changing the size of the molecule. QD size can be controlled during synthesis by adjusting the temperature, pH or feed-ratio of reagents. Another way to control the band gap energy is to modify the surface ligands attached to the surface of the quantum dot. Surface ligands are similar to dopants in semiconductors in that they alter the band gap energy by providing additional electrons or holes to the nanoparticle. The choice of surface ligand should also take into account what type of solvent will be used to dissolve the quantum dots, because the polarity of the ligand dictates whether particles will coalesce or dissolve in a polar or non-polar solvent. For example, Mercaptopropionic acid (MPA) is used as a surface ligand for water-soluble quantum
dots because it is a polar molecule and water is a polar solvent. The water-soluble CdTe quantum dots used in this study used a variety of mercaptocarboxylic acids as surface ligands with a basic structure of HS-R-COOH.

In addition to having very tunable electronic and optical properties, quantum dots have several desirable characteristics including broad absorption spectra, narrow fluorescence spectra, good photostability, fast decay time, no afterglow, high proton number, and high density. The broad absorption and narrow emission ranges can be explained by the electron energy transitions that are allowed by quantum mechanics, which I will not go into very deeply. The allowed energy transitions include excitations from the ground state to an excited state, any additional energy from the absorbed photon goes to higher vibrational and rotational energy sub-levels in the excited state as shown in Figure 1-2(B). However, fluorescence can only occur from the lowest sub-level of the excited state. This means an electron has to relax, through vibrational losses to the crystal lattice, before it can fluoresce. The difference in energy from the absorbed photon to emitted photon caused by relaxation is called the Stokes shift and can be seen in Figure 1-2(A). The requirement for an electron to relax and fluoresce from the lowest vibrational energy level is what makes the fluorescence spectrum narrower than the absorption spectrum. The width of the absorption spectrum would be equal to the height of S1 + S0 from Figure 1-2(B) and the fluorescence spectrum would be as wide as the height of S0. Fast decay times are due to the very quick relaxation times, which are on the order of 10^-8 seconds from the time of absorption to time of emission.

**Applications of Quantum Dots**

Potential applications of quantum dots include use in almost any type of optical device; in fact they are already being used in lasers, light emitting diodes (LEDs), and
solar cells. The major use of quantum dots at this time is in biomolecular imaging, through tagging of organic molecules. Quantum dots are very useful in this application because of their narrow fluorescence spectra in the visible range, which allows the use of several different quantum dot sizes to simultaneously image different molecules. Tagging is done by using surface ligands that bind to organic molecules; this allows in vivo activity to be monitored by exciting the dots with a UV lamp (or other photon source of appropriate wavelength) and observing the fluorescence using an optical microscope. The high quantum yield, photostability, and increased brightness compared to other currently used fluorescent biomarkers make quantum dots a better option than others. Future applications of quantum dots may include use in quantum computing, and x-ray imaging phosphor screens. Quantum dots may be useful as phosphor screens because of their fast decay times, lack of afterglow, high-Z, and fluorescence spectra. The function of an x-ray phosphor screen is to convert incoming x-rays into visible light, the visible light then is either detected by a charge coupled device (CCD) or interacts with a film emulsion. The high-Z of quantum dots is a desirable quality in phosphor screens because of the higher rate of x-ray absorption. This means that an image can be obtained with less flux, and consequently lower dose, than would be necessary with a phosphor screen of low-Z material. Fast decay times and lack of afterglow may improve spatial resolution in imaging techniques where the detector is in motion, such as in computed tomography (CT), because the image is produced almost instantly and motion becomes insignificant. The fluorescence spectrum of many quantum dots make them appropriate for phosphor screens because their spectrums coincide well with the absorption spectra of CCDs, which detect the fluorescent light and create the image.
Just as the increased absorption of high-Z material improves detection efficiency, the overlapping spectra of the quantum dot fluorescence and CCD absorption also improves the efficiency and, thus, also helps decrease dose to the patient.\textsuperscript{14}

**CdTe Quantum Dots**

In this study, 600nm emission peak CdTe quantum dots (2.5:1 Cd to Te ratio) were used because of the properties of quantum dots previously described. Quantum dots have the potential to be used for in vivo dosimetry because of their wide absorption spectrum and their fluorescence in the visible range. By converting x-rays into visible light, dose can be determined by measuring the amount of fluorescence via a photomultiplier tube. The high Z of CdTe allows the detector to be made very small and near tissue equivalent because a low concentration of quantum dots that does not appreciably change the attenuation properties of water can still obtain a statistically significant number of x-ray interactions in the detector. Figures 1-3, 1-4, and 1-5 show the mass attenuation spectrums of CdTe quantum dots, water, and the quantum dot solution. By using a small detector and aqueous solution of CdTe, the x-ray fluence is minimally perturbed allowing for accurate in vivo measurement. The fast decay time and lack of afterglow makes it more accurate temporally as a real-time dosimeter, and the fluorescence spectrum makes detection via photomultiplier tube simple.

**Objectives**

This study’s main goals are to determine the feasibility of using quantum dots as the scintillating material in an FOC dosimeter and to characterize its energy dependence, linearity, dose rate dependence, and reproducibility. Despite recent investigations attempting to use quantum dots as x-ray phosphors in medical imaging, this study is the first known attempt at using quantum dots for the purpose of medical
dosimetry. The plan is to potentially use this new detector as a real-time in vivo dosimeter in diagnostic and therapeutic procedures. In diagnostic procedures the dosimeter would generally be placed in the imaging field but away from the anatomy of interest so to not interfere with the imaging process. In therapeutic procedures, as is already done with diodes, TLDs, and MOSFETs, the dosimeter would be placed on the skin to measure entrance and exit skin exposures.

The characterization process will be, in general, to expose the detector to different radiation fields and compare its measurements with a reference ion chamber’s measurements in the same field.

Figure 1-1. Representations of the different types of quantum dots. A) Core-shell quantum dot with portion of outer layer peeled away. B) Colloidal quantum dot. C) Epitaxial quantum dot bound to crystal substrate. Reprinted (adapted) with permission from:
(a) Evident Technologies.
(b) http://commons.wikimedia.org/wiki/File:QD-pyramide.JPG
(c) G. Han, T. Mokar, C. Ajo-Franklin, and B.E. Cohen, “Caged Quantum Dots,” American Chemical Society. (2008).
Figure 1-2. The following diagrams explain some of the optical properties of quantum dots. A) Diagram showing the stokes shift caused by relaxation. Shape and size of absorption/fluorescence curves are simplified to emphasize the Stokes shift. B) Jablonski diagram showing allowed energy level transitions.
Figure 1-3. Mass attenuation spectrum for CdTe in a ratio of 2.5 atoms of Cd to 1 atom of Te.
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Figure 1-4. Mass attenuation spectrum for pure water.

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M.J. Berger, J.H. Hubbell, S.M. Seltzer, J. Chang, J.S. Coursey, R. Sukumar,
CHAPTER 2
MATERIALS AND METHODS

Quantum Dots

The species used in this study were water soluble CdTe colloidal quantum dots with emission peak at a wavelength of 599nm and absorption local maximum at 567nm. These quantum dots and associated spectrums (Figure 2-1) were obtained from MK Impex Corporation. The CdTe quantum dots came in a dry powder, and made a clear red solution when dissolved in water. The ratio of Cd atoms to Te atoms was approximately 2.5:1.

Detector Assembly

The CdTe quantum dot FOC dosimeter was constructed using 6 main components; it consisted of optical fiber (plastic or glass), glass capillary tube, aqueous CdTe solution, epoxy based glue, reflective paint, and heat shrink tubing. First, one end of the capillary tube was sealed with epoxy and one end of the optical fiber. After allowing 24 hours for the glue to dry, the CdTe solution was injected via syringe into the other end of the capillary and topped off with a glue plug. These two steps were crucial in minimizing the amount of air trapped in the capillary, which could affect the coupling of the solution with the optical fiber. To minimize air bubbles, the syringe needle was carefully inserted as far as possible into the capillary tube and filled to the brim to minimize the gap between the top of solution and the glue plug. Another 24 hours later, a layer of glue was applied to the entire length of the capillary tube to strengthen the glass capillary and solidify the plug. Finally, a coat of white reflective paint was applied to maximize the number of photons entering the optical fiber. Figure 2-2 shows some of the stages of detector assembly.
After the detector construction is complete, it must be sealed light proof so that ambient light does not contaminate the signal. This is done by inserting the entire length of the optical fiber and detector element into heat shrink tubing and sealing any openings. Finally, an SMA connector is glued onto the opposite end of the optical fiber and the end is polished to allow connection to the PMT system.

In order to discount the Cerenkov radiation produced within the optical fiber from the overall signal, a blank optical fiber is inserted alongside the detector fiber inside the heat shrink tubing. The blank fiber has its own SMA connector, so that the Cerenkov photons can also be measured by a PMT. This signal is subtracted from the detector fiber signal in order to obtain the portion of the signal that is due to quantum dot fluorescence.

A total of three detectors were built and characterized in this study. The first two, which we will refer to as 1a and 1b, were built as described above using 1,000µm diameter silica glass optical fiber. However, after inserting them into their respective heat shrink tubing they were inserted parallel to each other into larger diameter tubing so that when exposed they would be exposed to the same radiation and geometric conditions, and thus their characteristics could be more accurately compared. Finally, detector 2 was built using 510µm diameter plastic optical fiber. The glass capillary tubes used had inner diameters of 1,100µm and were filled approximately 1cm lengthwise with solution, giving a total active volume of approximately 0.01cm$^3$.

**Measurement Setups**

Most measurements had the same basic setup, the sealed detector was placed in a radiation field alongside an ion chamber used to obtain a reference measurement. A kV meter was also necessary in some cases to function as a trigger for the ion
chamber to record exposure. The major differences between experiments came in the energy or mAs used. The different radiation sources used throughout this study were an Alpha III MGF-110 mammography unit (20-35kVp), a Source-Ray SR-115-S x-ray tube (50-100kVp), and an Elekta Precise linear accelerator.

**Energy Dependence**

The quantum dot fluorescence dependence on energy was quantified by varying the kVp while keeping a constant mAs (or MU), this was done on the mammography, x-ray tube, and linear accelerator units. A reference chamber was used to provide a measure of the air kerma during exposures so that the energy dependence could be expressed in counts/mGy at a given energy. This allows the variations in fluence at different energies to be removed, giving a true measure of the dependence on energy rather than energy and flux.

**Mammography unit (20-35kVp)**

The mammography unit was used to measure the energy dependence at 20-35 kVp. The tube current was kept constant at 12 mAs throughout the energy range. Reference air kerma measurements were taken using a Keithley 96035B parallel plate ion chamber and Keithley 35050A electrometer. KVP corrections were done using the Keithley 35080B kVp divider, which also functioned as the trigger for the ion chamber.

**X-ray tube (50-100kVp)**

In effort to keep the results as similar as possible between x-ray sources, the tube current was again held constant at 12 mAs for energy ranges from 50-100 kVp on the x-ray tube. Again, the Keithley chamber, kVp divider, and electrometer were used for reference measurement. The setup used can be seen in Figure 2-3.
Linear accelerator (4, 6, 18MV)

Finally, the energy dependence was measured at 4, 6, and 18 MV on the linear accelerator. A parallel plate ion chamber was used for reference measurements taken at 5cm and 10 cm depth in solid water because ion chambers are not accurate in air (without buildup) at megavoltage photon energies. Unlike the low energy measurements, the reference measurements were not taken simultaneously with the CdTe detector measurements because both detectors could not fit in the solid water opening at the same time. The schematic in Figure 2-4 shows the setup used with the detectors and solid water. The field size was 10X10 at SPD=100, where SPD was measured to the effective point of measurement of the parallel plate chamber (top face of the chamber). This setup means that approximately 5cm of the optical fiber was in the primary beam and therefore highly susceptible to radiation induced attenuation (RIA), which will be explained later on in Chapter 4.

Linearity

The linearity was measured using similar setups as the energy dependence measurements, but did not require reference measurements for normalization. Instead of varying the energy, the mAs or MU was varied.

Mammography unit

The same setup was used as in the energy dependence measurements, but the energy was held constant at 30kVp and the tube current was varied from 4-175 mAs.

X-ray tube

The same setup was used, while the energy was held constant at 75 kVp and the tube current varied from 3-60 mAs.
**Linear accelerator**

The linearity measurements on the linear accelerator were done a little differently. Since the linearity can be measured without exposure normalization, there was no reference chamber or solid water buildup used. The energy was kept at 6MV and the monitor units were varied from 50-350 MU. As in the energy dependence measurements, the detector was placed in the center of a 10X10 field, therefore the optical fiber was again susceptible to RIA.

**Dose Rate Dependence**

The dose rate dependence was measured using a similar setup as shown in Figure 2-3. Exposures were done on the x-ray tube at 90 kVp and 30mAs. The dose rate was varied by changing the distance from the tube to the detectors, thereby taking advantage of the divergence of the beam to reduce the energy fluence as the distance increased. Again, the Keithley chamber, kVp divider, and electrometer were used for reference measurement.

**Reproducibility**

The reproducibility was measured by taking 30 of the same measurements with identical setups. The setup was taken down and redone after every 3 measurements. This was done because air bubbles moving within the capillary tube can affect coupling with the optical fiber, and therefore can affect the number of counts registered. The coefficient of variation (COV) between setups was found to quantify the reproducibility. Since the reproducibility is more a function of the detection system and not the radiation parameters, the reproducibility was only measured with the x-ray tube at constant settings of 75kVp and 24mAs.
Detectors used

Measurements were taken at different times with detectors 1(a, b) and 2. All were used for measurements on the x-ray tube and the mammography unit, but the optical fibers in 1a and 1b were too short for use in the linear accelerator measurements because the PMT system had to reach outside of the treatment room. All measurements described, except the linear accelerator and dose rate dependence measurements, were done with detectors 1a and 1b. Detector 2 was used for all measurements described in Chapter 2.

Figure 2-1. Properties of CdTe quantum dots used in this study. A) Fluorescence spectrum shows relative intensities of fluorescence emissions vs wavelength. B) Absorption spectrum shows absorbed fraction of incident radiation vs wavelength.
Figure 2-2. Stages of detector assembly. A) Right after the epoxy plug is placed. B) After application of the layer of glue. C) Reflective paint layer. (Photos courtesy of Juan Tellez.)

Figure 2-3. Different views of a typical setup on the portable X-ray unit. A) A view of the x-ray tube directed at the detectors and kV meter. B) A view of the detectors from the x-ray tube’s point of view. (Photos courtesy of Juan Tellez.)
Figure 2-4. Schematic showing the solid water setup used in energy dependence measurements on the linear accelerator.
CHAPTER 3
RESULTS

Energy Dependence

X-ray Tube and Mammography Unit

In the mammography energy ranges, the energy dependence appeared to have a logarithmic shape with the slope plateauing as the kVp increased. However, in the x-ray energies ranging from 50-100 kVp, the energy dependence behaved linearly. In both ranges, 1a and 2 behaved very similarly while 1b was more of an outlier as can be seen in Figure 3-1.

Linear Accelerator

Unlike the energy dependence measurements taken on the mammography unit and the x-ray tube, the linear accelerator measurements actually showed a decrease in normalized counts with increasing energy. As shown in Figure 3-2, the data points appear to decrease in an exponential manner rather than increasing in a linear fashion as occurred at diagnostic energies shown in Figure 3-1.

Linearity

Mammography Unit

Although each detector displayed different count rates, evidenced by different slopes in Figure 3-3, all three detectors showed very good dose linearity.

X-ray Tube

Again, all detectors showed very good linearity. Despite a few points deviating from the fitted line for detector 1b, the data was still accurately represented by a linear fit. Although detector 1a and 2 had different slopes (count rates) when measured on the
mammography unit, their slopes were nearly identical when measured on the x-ray tube.

**Linear Accelerator**

Even at much higher count rates than at diagnostic energies, the measurements on the linear accelerator showed very good dose linearity. Only detector 2 was used in this measurement because the length of the optical fiber was not long enough on detectors 1a and 1b.

**Dose Rate Dependence**

The normalized counts remained relatively constant (~45 counts/mGy) throughout the range of dose rates measured by detector 2. Counts were normalized by air kerma to remove any dependence on total dose, so that the dose rate dependence can be extracted.

**Reproducibility**

The reproducibility was measured by repeating the same measurement over and over, with the setup repeated 10 times and 3 measurements made per setup. The detectors did not perform as well as desired, with only detector 2 having a coefficient of variation below 5%. Again, the counts were normalized to remove dependence on total dose, so that the reproducibility of the detectors could be extracted.
Figure 3-1. Energy dependence measurements from all three detectors on the mammography unit (20-35kVp) and x-ray tube (50-100kVp) compiled into one graph. Constant tube current of 12mAs used throughout.
Figure 3-2. Energy dependence measured at 4, 6, and 18 MV on an Elekta Precise linear accelerator.
Figure 3-3. Linearity measurement done on the mammography unit at 35kVp.
Figure 3-4. Linearity measurements done on the x-ray tube at 75kVp.
Figure 3-5. Linearity measurements done on the Elekta Precise linear accelerator at 6MV.
Figure 3-6. Dose rate dependence measured on the x-ray tube at 90kVp and 30mA.

Counts/mGy

mGy/ms
Figure 3-7. Reproducibility as measured on the x-ray tube with constant parameters of 75kVp and 24mAs.
CHAPTER 4
DISCUSSION

Energy dependence

The energy dependence of a radiation detector is a measure of how the efficiency of a detector in detecting incident radiation changes with the energy of the incident radiation. The energy dependence of FOC dosimeters can be quantified by how many light photon counts the PMT measures per unit air kerma as a function of energy. However, other factors such as geometry and beam quality can play a factor in measuring the energy dependence. For example, in Figure 3-1 there is an apparent discontinuity between the mammography and x-ray tube ranges, this is likely due to different scatter conditions and different beam qualities.

Mammography Unit and x-ray Tube

It was expected that no two detectors would behave exactly the same because of differences in active volume and/or quality of coupling to the optical fiber. However, since the detecting material was the same for all detectors, they were expected to have similar slopes (energy dependence). It is difficult to say why detector 1b had such a large slope compared to detectors 1a and 2; on one hand detector 1b was an outlier, but on the other hand it had by far the highest count rate and therefore was statistically more reliable than the other two detectors. Table 4-1 gives the equations and coefficients of determination ($R^2$) for the best fit curves of all detectors in the mammography and x-ray tube ranges.

A possible explanation for the differences between detectors is that the quality of coupling affects the magnitude of counts, while the size of the active volume affects the slope as well as the magnitude. The larger active volume would allow for higher count
rates (larger slope) due to more available quantum dots to potentially absorb an x-ray. So, from the data collected in this study, it is impossible to designate an energy dependence of CdTe quantum dots in general because the energy dependence seems to be specific to individual detectors based on their active volume size and quality of coupling.

**Linear Accelerator**

The energy dependence at megavoltage energies did not behave as expected. A best fit exponential curve had the formula \( (3 \times 10^6) e^{-0.089kVp} \) with an \( R^2 = 0.950 \). Unlike at diagnostic energies, the normalized detector counts actually decreased as the energy was increased. This could be for a number of reasons. One possible reason is that the reference ion chamber used for normalization may not be appropriate for this situation. Solid water was used for buildup so that the ion chamber could measure a stable signal, but ion chambers measure charge created through ionizations while the CdTe FOC dosimeter measures how much absorption/fluorescence is occurring. Therefore the use of buildup will increase the measurement made by an ion chamber more than it will for the CdTe FOC dosimeter. Another, less likely, possibility is that the higher energies are simply penetrating through the detector and fewer counts are being registered. This is unlikely because the mass attenuation coefficients do not change nearly enough to account for the decrease in counts. A final possibility is that the RIA due to the optical fiber being exposed to very high dose rates characteristic of pulsed radiation from the linear accelerator is greater at higher energies; RIA will be described in the Dose Rate Dependence subsection of this chapter in further detail.
**Linearity**

When choosing a material for the sensitive element of a radiation dosimeter, a desirable characteristic is that it have a linear dose response. A linear response exhibits constant count rate throughout an exposure regardless of the cumulative dose. The coefficients of determination in Table 4-2 for the linear fits all come out nearly equal to 1. This means that the data is very accurately represented by a linear model, and thus shows that the detector has a linear dose response.

An important characteristic of this CdTe FOC dosimeter that allows it to behave linearly is that it functions as a real time dosimeter. As a real time dosimeter, FOC dosimeters do not accumulate excited electrons in the way dose integrators such as TLDs do. Accumulation occurs in TLDs because excited electrons are held in traps until after the exposure is complete, then the electrons are stimulated back to the ground state in order to measure the dose delivered. By accumulating excited electrons, the dose response of TLDs begins to decline after they have been exposed to a certain amount of dose. However, this decline does not happen until very high doses (~100Gy for LiF TLDs) are met and are not clinically realistic. Accumulation does not occur in CdTe FOC dosimeters because of the fast decay time (~$10^{-8}$ sec) of the quantum dots. Almost immediately after being excited, the quantum dots relax and fluoresce. After fluorescing, they are again available for excitation. This fast decay process allows the dose response to remain constant because there are always quantum dots available to be excited. Only at very high dose rates will a CdTe FOC dosimeter potentially lose dose response and become saturated.

It is important to note that although there were no issues with RIA in the linearity measurements, it was still present. Since the linearity is measured at a constant energy
and dose rate, the RIA was constant throughout and therefore did not pose any issues with data analysis.

**Dose Rate Dependence**

Detector 2 did not display strong dose rate dependence, at least at the measured diagnostic dose rates. There was a slight increase in normalized counts as the dose rate increased, but the counts hovered around 45 counts/mGy throughout the dose rate range measured. The counting statistics were not large enough to say for certain that the normalized counts actually increase with dose rate. A line fitted to the data with slope of 118.34 had an $R^2$ value of only 0.71044, which does not give much strength to the counts increase with dose rate argument. However, the consistency of the counts hovering around 45 counts/mGy suggests that at typical diagnostic dose rates the dependence of counts on dose rate is negligible, if not zero.

The dose rate was not analyzed at therapeutic energies, but if quantum dots have any dose rate dependence at all, it is likely to be observed at these much higher rates. A potential issue when trying to measure dose rate dependence at therapeutic energies is that fiber optics have been shown to have changes in light attenuation characteristics when exposed to pulsed radiation (linear accelerators produce pulsed x-rays). This is because of so called “transient color centers” that are created that change the fiber’s light transmitting properties.\(^{15}\) This radiation induced attenuation (RIA) of the fiber optic may be the reason behind the decreased count rate in the megavoltage energy dependence measurements seen in Figure 3-2. This phenomenon of RIA adds an element of complexity to characterizing the dose rate dependence because the system may now have a dose rate dependence of the detecting element as well as the light-transmitting element.
Reproducibility

Dosimeter reproducibility is very important because if a repeated measurement under the same conditions results in a different value every time, it is impossible to know which result, if any, is accurate. Under identical conditions, a good dosimeter should give steady results within reasonable statistical variation given the nature of radiation detectors depends on statistical processes. The most consistent detector tested achieved a coefficient of variation (COV) of 2.27%, which is comparable to the low end of other commonly used in vivo dosimetry systems. Table 4-3 shows the coefficients of variation for several different in vivo dosimeters. However, note that these tabulated values for TLD, diodes, and MOSFETS were found at therapeutic energies, while the CdTe FOC dosimeter values were found at 75 kVp. Reproducibility values can vary at different energies for the same dosimeter; MOSFETs, for example, can have reproducibilities ranging from 0.155-0.318 in the mammography range compared to only 0.015 at therapeutic energies.\textsuperscript{16, 17} So, CdTe FOC dosimeters compare favorably to MOSFET at diagnostic energies, but still to be determined for the therapeutic range.

Possible contributors to detector instability include air bubbles in the capillary tube that can alter the quality of coupling just by moving around, or even coils in the fiber optic cable that can allow light to escape through bends and alter the efficiency of light transmission from the active volume to the PMT. Typically, the reproducibility would be measured while using the same setup rather than taking it down and reassembling the apparatus, but for the reasons just described, the setup was repeated over and over to take into account the variations in the coupling quality and efficiency of light transmission of the detecting system. Otherwise, repeating the setup introduces additional variation into the reproducibility due to the inability of the operator to precisely
recreate the setup the same as before. If the issues of coupling quality and light transmission are addressed, there is no reason to believe a CdTe FOC dosimeter could not achieve a COV of under 2% which may be acceptable for therapeutic use.

Table 4-1. Energy dependence of each detector in the mammography and x-ray tube ranges, given by best fit curve.

<table>
<thead>
<tr>
<th>Detector</th>
<th>Mammography range best fit curve</th>
<th>R²</th>
<th>x-ray tube best fit curve</th>
<th>R²</th>
</tr>
</thead>
<tbody>
<tr>
<td>1a</td>
<td>-2.599kVp² +208.37kVp-1778.4</td>
<td>0.995</td>
<td>20.87*kVp+4768</td>
<td>0.955</td>
</tr>
<tr>
<td>1b</td>
<td>-4.880kVp² +386.49kVp-3330.6</td>
<td>0.995</td>
<td>94.62*kVp +3956</td>
<td>0.985</td>
</tr>
<tr>
<td>2</td>
<td>-.0.551kVp² +62.53kVp-112.59</td>
<td>0.978</td>
<td>15.04*kVp +4061</td>
<td>0.927</td>
</tr>
</tbody>
</table>

Table 4-2. Slope and coefficients of determination (R²) for linear fits of each detector.

<table>
<thead>
<tr>
<th>Detector</th>
<th>Mammography unit Slope</th>
<th>R²</th>
<th>x-ray tube Slope</th>
<th>R²</th>
<th>Linear accelerator Slope</th>
<th>R²</th>
</tr>
</thead>
<tbody>
<tr>
<td>1a</td>
<td>323.2</td>
<td>0.99997</td>
<td>655.3</td>
<td>0.99986</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>1b</td>
<td>555.4</td>
<td>0.99998</td>
<td>1419.1</td>
<td>0.99244</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>2</td>
<td>205.1</td>
<td>0.99997</td>
<td>653.3</td>
<td>0.99996</td>
<td>1173.9</td>
<td>0.9996</td>
</tr>
</tbody>
</table>

Table 4-3. Coefficients of variation for CdTe FOC dosimeter and other commonly used in vivo dosimeters.

<table>
<thead>
<tr>
<th>Detector</th>
<th>COV</th>
</tr>
</thead>
<tbody>
<tr>
<td>1a</td>
<td>0.1467</td>
</tr>
<tr>
<td>1b</td>
<td>0.0774</td>
</tr>
<tr>
<td>2</td>
<td>0.0227</td>
</tr>
<tr>
<td>TLD¹⁸</td>
<td>0.006-0.0327</td>
</tr>
<tr>
<td>Diode⁶</td>
<td>&lt;0.01</td>
</tr>
<tr>
<td>MOSFET¹⁷</td>
<td>0.015</td>
</tr>
</tbody>
</table>
CHAPTER 5
CONCLUSION

A well-made quantum dot FOC dosimeter can be made to be fairly accurate. In this study, water-soluble CdTe quantum dot FOC dosimeters with absorption peak of 567nm and fluorescence peak of 599nm were characterized for their energy dependence, linearity, dose rate dependence, and system reproducibility. Each of the three detectors had slightly different values for the four parameters measured, this was thought to be due to quality of coupling and size of active volume. All three detectors gave reasonably smooth curves for the energy dependence and linearity measurements, but detector 2 drastically outperformed the other two in the reproducibility measurements with a COV of only 0.0227 compared to 0.1467 and 0.0774 for detectors 1a and 1b respectively. If this type of FOC dosimeter were to be used clinically, it would have to be calibrated individually for each machine because different machines, especially with different x-ray generating target materials, produce different energy spectra and therefore would have different energy dependence.

Suspension of the quantum dots in a plastic PMMA matrix, rather than using an aqueous solution, was briefly investigated but ultimately decided against. In the future, the possibility of using PMMA should be investigated further to assess whether or not coupling can be improved by using solid PMMA to avoid the air bubbles in solution issue. Using PMMA to get rid of the air bubbles would likely improve the reproducibility of the system, possibly enough to use the dosimeter clinically in the most demanding settings. Even with improved reproducibility, the relevant details of RIA would have to be better understood before using fiber optics to measure dose from pulsed radiation.
Although the reproducibility was not directly measured for therapeutic energies, it is fair to say that it would improve compared to that of the diagnostic range. This can be inferred due to the higher counting statistics at higher energies and the fact that the reproducibility of TLDs and MOSFETs decreases in the mammography range compared to the therapeutic range.\textsuperscript{16, 19} If the CdTe FOC dosimeter does indeed follow the pattern of TLDs and MOSFETs, it is very possible that it would behave well enough to be used in a clinical setting given that it would come close to the AAPM standard for diode reproducibility for in vivo dosimetry in radiation therapy. Given that the CdTe FOC dosimeter is composed of near tissue equivalent material, it would be an excellent option for an in vivo dosimeter because it would be able to measure parameters such as energy and flux while causing minimal perturbations. Potential use in the clinic would be contingent on confirmation that the reproducibility does in fact improve at therapeutic energies and that the RIA of fiber optics is better understood and can be accounted for. That being said, even at this stage, the CdTe FOC dosimeter studied may be the best option for in vivo dosimetry at diagnostic energies given that it has good linearity, characterizable energy dependence, negligible dose rate dependence, and better reproducibility than TLDs, diodes, and MOSFETs.
LIST OF REFERENCES


BIOGRAPHICAL SKETCH

Juan Manuel Tellez was born in Toronto, Canada to Camilo and Stella Tellez. He is the second of two children, after his older sister Beatriz. After moving to Michigan in his early childhood, Juan eventually graduated from Wylie E. Groves High School in Beverly Hills, Michigan in 2007. He then went to the University of Illinois to pursue his bachelor’s degree in Engineering Physics where he first came across the field of Medical Physics. His desire to become a Medical Physicist then brought him to the University of Florida where he will be finishing his master’s degree in Biomedical Engineering in August 2013.